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“HIGH SUSTAINED G-TOLERANCE MODEL DEVELOPMENT”

FINAL REPORT

Project Manager

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INTRODUCTION

The basic mathematical model (BMM), realized as special computer based information technology (SCIT) and named “PILACCEL”, was developed during the previous research and described in [17]. It was able to provide simulation investigations in area of human acceleration physiology. The main function of the basic SCIT was to simulate the hemodynamics of the “mean” man under piloting Gz accelerations.

The main goal of this development phase is to improve previous development and to create the general concept, models, algorithms and software that will include the following additional simulation opportunities:

1. The models have to be tunable by user to the individuals including their sex;
2. A special model of the “Libelle” anti-G suit will be developed.
3. The user manual will be created;
4. Special recommendations concerning optimal protections will be generated.

The final report presented here describes the final modification of the basic SCIT and its transformation to the actual “PILACCEL”. Now the “PILACCEL” is able to provide all the functions that were planned to be created. During this development phase was essentially changed the basic model presented in [17]. Therefore, the final report includes in it not the description of the new part only, but all the actual models and the actual user interface. This report also consists of an analysis concerning problems included above in tasks 2 and 4. Besides, the report presents the results of our several additional efforts that were not planned, but they arose during the development. Special efforts were directed to change and to develop the software interface so, to do it more friendly to the user-physiologist. The final “PILACCEL” includes a “Help”-module in its with a short description of the theoretical mathematical models and the user manual.

The theoretical basis of the “PILACCEL” software is a special complex of the quantitative mathematical models. This complex consists of two different blocks. The first block of models, named as physiologic model (PM), describes the main physiological regularities that determine the human reactivity to the changes of the inertial forces. It mainly presents the behavior of the human cardiovascular system under high sustained and dynamic Gz accelerations. The second block of the models, named as environmental model (EM), has not any direct relation to the human physiology. But it presents all the outside forces, usually generating to modulate the human physiology, and to resist of arterial pressure such extremely decreasing, that origins a central light loss (CLL). These outside forces are the specific environment, in which will function the pilot under the extreme accelerations. This environment includes several additional natural and/or artificial procedures, specially generating to increase the human tolerance to the Gz accelerations.

PART 1:

THEORETICAL BASIS OF THE “PILACCEL” SOFTWARE

1 MATHEMATICAL MODELS OF THE HUMAN PHYSIOLOGY

According to the general demands to the PM, it will be able to reflect the human hemodynamics both for the mean men and also for the individuals. In the last case the PM will include the gender aspect too.

1.1 General concept of the human physiology modeling under Gz accelerations: basic and individual models.

We had two starting conditions for creating the PM. The first one is that the basic mathematical model complex (BMMC) will be created. The BMMC must be able to describe almost all known physiologic mechanisms that may be involved with the healthy mean man organism in general activation processes to reach some level of human tolerance to dynamic high sustained +Gz accelerations. According to the second condition, the required models must be adaptable by the user to the persons under the same situations. To reflect the main individual variations of person's hemodynamics, models will include several additional characteristics. These data, concerning persons' sex, weight, age, body length and. health, have to be set by the user during his/her interaction with the “PILACCEL”. These models will be presented as an appropriate research technology oriented towards the area of human acceleration physiology.

The BMMC is a special problem-oriented model complex. Every personal model has to be created automatically by relatively simple procedures, according to the actual input data.

The BMMC includes several models:

- the model of heart pump function (HPFM);
- the model of the systemic and lung vascular hemodynamics (VHM);
- the model of the baroreflexor control in cardiovascular system (BCM);
- the model of the additional reflexes (ARM), involving under high sustained Gz accelerations.

The BMMC was created in frame of well known modeling approach that allows to present the real cardiovascular system as a net of vascular compartments, located in different levels and environments (cavities, tissues), but connected through hydraulic resistances.

1.2 Model of heart pump function.

The HPFM discloses main relationships between the mean values of cardiac output (Q) and central venous (P_v - for right heart) or lung venous (P_{lv} - for left heart) pressures. Additional factors that have been taken into consideration, are the heart rate (F) and the inotropic coefficient (k) of ventriculae, the resistance (R_{av}) of atria-ventricular valves, the duration of diastole (T), the diastolic rigidity (D), and the unstressed volume (U) of ventriculae:

$$\begin{aligned}
Q &= \frac{F \cdot k_i \cdot [(\Delta P_{ai}^V \cdot / D_i + U_i) - U_o] M}{[1 - (1 - k_i)] \cdot M} \\
M &= 1 - \exp\left(-\frac{T_L D_i}{R_{aVi}^K}\right) \\
P_{ai}^V &= P_i^{INP} + 0.735 * r * H_a^V * N^G * \sin j - P_i^{FD} \\
k &= k_0 / [1.0 + 0.005 * (P_A - 160)], P_A > 160 \\
T_L &= \frac{1}{F} \cdot A + B \cdot (1 - k) * (1 - 0.1 * (a/20)), \\
D_i &= D_{i0} * [1 + 0.06 * (V_i - V_{1i})], V_i > V_{1i} \quad i = 1, 3 \\
D_i &= D_{i0} * [1 + 0.01 * (P_g - 5)], 5 \leq P_g \leq 125, i = 2, 4 \\
P_i &= \begin{cases} 0, V_i < U_i \\ (U_{1i} - U_i) * D_i, U_i \leq V_i \leq U_{1i} \\ (U_{1i} - U_i) * D_i + (V_i - U_{1i}) * D_{1i}, V_i > U_{1i} \end{cases} \\
P_i &= 2.5 * U_i * D_i + 0.002 * (V_i - 3.5 * U_i)^2, V_i > 3.5 * U_i, i = 1, 3 \\
V_i(t) &= V_i(0) + \int_0^t (Q^I - Q^O) dt \\
V^s &= \begin{cases} k_0 V^{ED} - V_0, P_v \leq P_0 \\ k_1 V^{ED} - V_1, P_v > P_0 \end{cases}
\end{aligned} \tag{1.1}$$

Symbols used for the model of heart pump function:

- $i = 1, 3$ - low index that defined property to the right and left atrium of heart
- $i = 2, 4$ - low index that defined property to the right and left ventricle of heart
- H_a^V - distance between atrial and ventricular centers
- N^G - coefficient of gravitational overload
- j - gravitational vector deflexion angle
- P_i^{FD} - end - diastolic pressure
- r, A, B - approximation constants
- Q^i, Q^o - input and output flows respectively
- r_1, r_2 - resistances of opened and closed valve respectively
- ΔP_k - pressure difference on both valve sides
- P_{KP} - pressure necessary for valve opening
- P_{KO} - pressure necessary for valve closing
- T_L - duration of heart diastolic phase
- D - diastolic rigidity of ventricle
- V - strained volume
- a - person's age
- F - heart rate
- k - inotropic coefficient of ventricle.

The last formula in (1.1) is another reflection of the well-known regularity of HPF. This regularity is also known as the heterometric or Frank-Starling's mechanism of self-control of heart function. It shows that relations between stroke volume (V_i^s) of ventriculae and their end-diastolic volume (V_i^{ed}) may be presented as a linear approximation. The coefficient k characterizes the level of inotropism of heart. Therefore, using k we can establish relations between influence of concentration of cardioactive humoral agents or rate of efferent nervous activity, and the inotropic state on the right or left ventriculae of heart.

1.3 Mathematical model of the systemic and lung hemodynamics.

The biophysical pressure-volume (P-V) static characteristics of different arterial or venous vessels are essentially nonlinear and different for different arterial or venous vessels. These nonlinear curves in each i -th compartment of vessels are approximated in the model by means of piecewise-linear characteristics, consist of three parts. According to this approximation, a typical description of P-V- dependency looks like:

$$P_i^T = \begin{cases} (V_i - U_i) \cdot D_{0i} & , V_i < U_i \\ (V_i - U_i) \cdot D_{li} & , U_i \leq V_i \leq U_{li} \\ (U_{li} - U_i) \cdot D_{2i} + (V_i - U_i) \cdot D_{li} & , V_i > U_{li} \end{cases} \quad (1.2)$$

Blood flows between j -th and l -th vessel compartments, which are connected by means of hydraulic resistance R_{jl} , are defined as a result of division of pressure gradients (G_{jl}^P) by R_{jl} . Transmural pressures, external pressures P^E and hydrostatic pressures P^G are considered as factors in determining G_{jl}^P . Coefficients for P^E reflect differences in the levels of vessels' location and transmission characteristics of different vessels environment (muscles, cavities, skin):

The modeling of hemodynamic effects of acceleration is based on the calculation of every hydrostatic pressure as a function of human posture and the value of acceleration. We have two compartment level classes. The first characterizes the value of distance between the human feet and place of localization of vessel compartment for human clinostatic or erect positions. The second class (we call it real levels) reflects the value of hydrostatic pressures of human vessel compartments for all other his/her positions. Using angle values between the horizontal and the directions of different body parts (α – for calf, β – for thigh, and γ – for all other compartments of body and head vessels), these parameters can be calculated in the model according to the following formulae:

$$\begin{aligned} L^s &= 0.5 \times A \times l^s \times \sin \mathbf{a} \\ L_1^t &= (L_c - 0.5 \times l_1^t \times \sin \mathbf{b}) \times A \\ L_2^t &= L_1^t - 0.5 \times A \times l_2^t \times \sin \mathbf{b} \\ L_p &= L_c \times A \\ L_{li}^b &= (l_{li}^b - L_0) \times A \times \sin \mathbf{g} + L_0 \end{aligned} \quad (1.3)$$

where l^s – length of calf, l_1^t – l_2^t – lengths of two parts of thigh, L_c – summary length of legs, L^s – level of shank vessel compartment, L_1^t and L_2^t – real levels of thigh vessels compartments, L_p – level of aviation armchair seat place, L_0 , L_{li}^b , l_{li}^b – real level and initial length of localization for each i -th body or head vessel compartment.

$$\begin{aligned}
q_{jl} &= \frac{G_{jl}^P}{R_{jl}} \\
G_{jl}^P &= (P_j^T + K_j^e \cdot P_j^E) - (P_l^T + K_l^e \cdot P_l^E) + P_{jl}^G \\
P_{jl}^G &= N^G \cdot \mathbf{r} \cdot (L_j - L_l)
\end{aligned} \tag{1.4}$$

For collapsible vessels we use special formulae to define the values of their resistances:

$$R_1 = \begin{cases} R_0 \cdot \left(\frac{V_0}{V} \right)^2, P^T > P_0 \\ R_0 \cdot r_0^4 \cdot \frac{a^2 + b^2}{2a^3 \cdot b^3}, P_1 \leq P^T \leq P_0 \\ R_1, R_1 \gg R_0, P_1 < P_0 \end{cases}$$

$$a = \frac{V \cdot r_0^2}{V_0 \cdot b}, \quad b = \frac{1}{3} r_0 \cdot \left[d + 2 \cdot \left(1 + \sqrt{1 - 2d^2 + d} \right) \right], \quad d = \frac{V}{V_0} \tag{1.5}$$

$$R = R_u \cdot \left(\frac{U}{V} \right)^2, \quad V_0 = V|_{p=0}$$

The behavior of summary brain flow depends on nervous origin resistance R^{AM} changes with time constant \mathbf{d}_M and pressure changes in cerebral liquid

$$R^{AM} = \begin{cases} R_{\min}^{AM}, P^{AM} \geq P_{\max}^{AM}, P_{\min}^{AM} < P^{AM} < P_{\min}^C \\ R_{\max}^{AM} \cdot C, P_{\max}^C < P^{AM} < P_{\max}^{AM} \\ \frac{E_1}{P^{AM}}, 0 \leq P^{AM} \leq P_{\min}^{AM} \end{cases}$$

where

$$C = [1 - \exp(X_i \cdot (P^{AM} - P_{\max}^{AM}))] \tag{1.6}$$

$$\frac{d R^{AM}(t)}{dt} = \frac{\mathbf{d}_M \cdot P^{AM}(t) - R^{AM}(t)}{T_m}, P_{\min}^C < P^{AM} < P_{\max}^C$$

The function of systemic venous valves is described in models as:

$$R_{jv} = \begin{cases} R_{1j}, q_j > 0 \\ R_{2j}, q_j \leq 0 \end{cases}, R_{1j} \gg R_{2j} \tag{1.7}$$

Dynamics of blood volumes in compartments are described by the following equations:

$$\begin{aligned}
V_j(t) &= V_j(0) + \int_0^t (q_j(t) - q_l(t)) dt \\
\sum_i V_i(t) &= \text{const}
\end{aligned} \tag{1.8}$$

According to the last equation, the summary blood volume in CVS is stable during the relatively short time interval of simulation experiment.

The basic VHM includes 27 systemic and 6 lung arterial and venous compartments located on the different levels. The heart is presented in a HPFM by its 4 chambers. Three environments (cranial, thoracic, abdominal cavities) are specially presented as extravascular environments with their specific conditions of extravascular pressure dynamics. So, in having such a presentation of vascular net, we are able to simulate important influences of extravascular pressure changes in these cavities on local hemodynamics. Having aortic arch and carotid sinus compartments, we will be able to describe relationships between afferent nervous activity and transmural pressures in these reflexogenic vascular areas important for systemic arterial baroreflex .

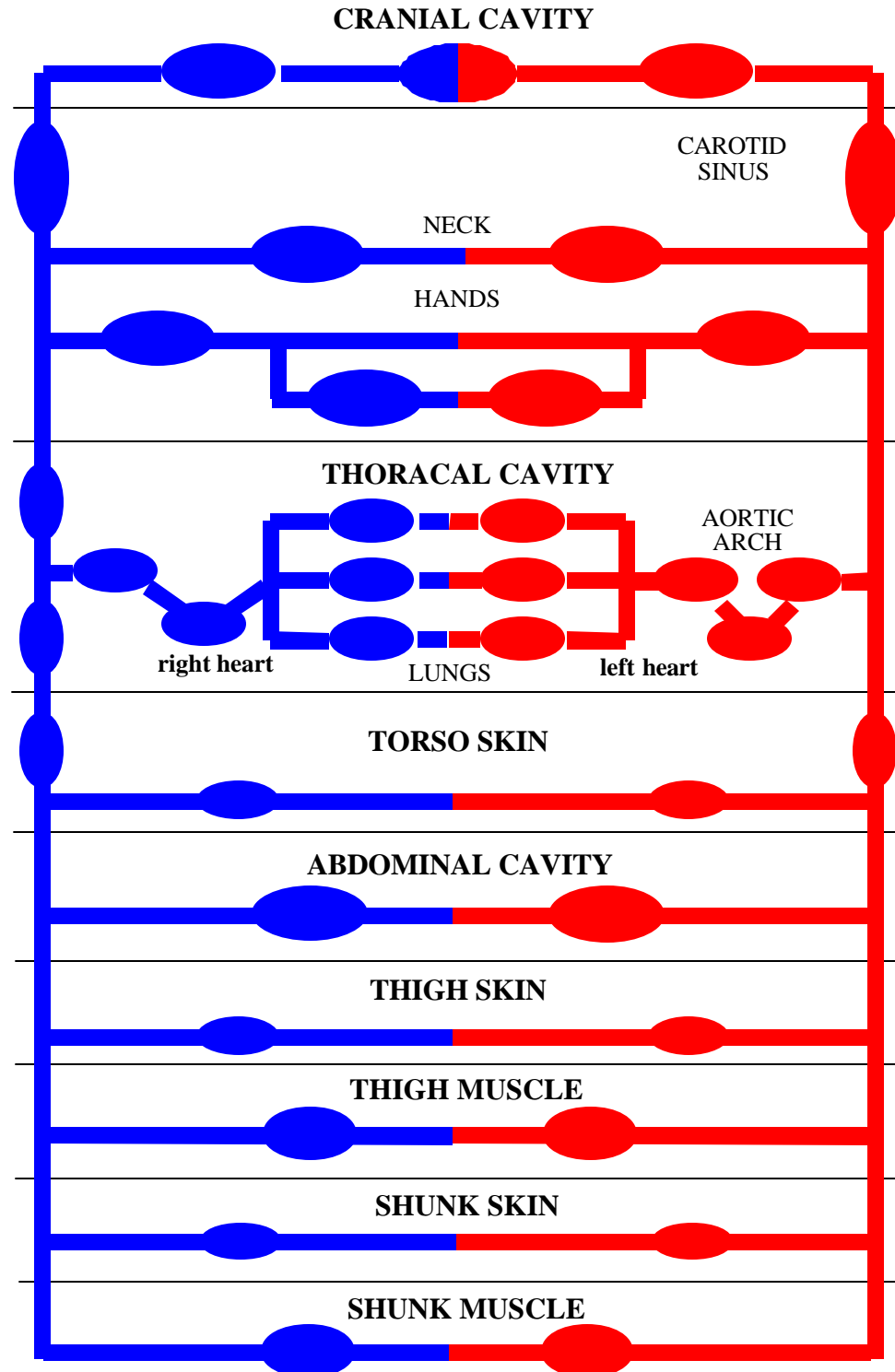


Fig.1. A presentation of multicompartmental CVS in the HPFM and VHM.

1.4 Mathematical model of the baroreflex control in cardiovascular system.

Symbols used for models of control mechanisms of hemodynamics:

- P^A –mean pressure in aortic arch
- P^C –mean pressure in carotid arteries
- P^{BA} –mean pressure in brain arteries
- j_j –angle between j compartment and horizontal
- h_j –length of compartment
- F_q^R –level of mechanoreceptor activity
- P_q^R –transmural pressure in reflexogenic area
- P_q^t –thresholds of receptors activity
- X_j –control parameters of CVS
- F_A, K_A –levels of F and K under automatic regime of heart
- a_i, b_i, p_j –constants
- d_1, d_1^u, b –constants
- P_{p-0}, P_p –intrapleural pressures under human supine and seat positions respectively

The general structure of the model of CVS's nervous control, consisting of BCM and ARM, is presented on Fig.2. There are two heart control parameters (F, k) and three integral control parameters of vessels (D, U, R). The last three parameters are dispersed on different regional areas of vessels, according to the existing physiological notions about their efferent sympathetic nervous density.

There are two different feedback channels on this scheme. The first one is divided in its turn, and forms two negative feedback channels for arterial baroreceptor reflexes (ABR) from baroreceptors of the aortic arch and the carotid sinus zones. The second feedback channel is a positive feedback channel for the mechanoreceptor reflex from the right atria area (this reflex is known as the Bainbridge reflex). The last one can only change F and k . The ABR is often included in different models of the hemodynamics control but the Bainbridge reflex is seldom presented in models. However, there are sufficient reasons to include it in our hemodynamics model under piloting accelerations. The first reason is the essential increasing of central venous pressure during special breathing procedures. The second reason is that the central venous pressure also increases during muscle stress or use of extended coverage anti-G-trousers.

The nervous activity in both feedback channels is formed by the difference between the set-points in the central neuronal structures level and the summary activity of receptors. ABR is presented in the model as proportional regulators. They are based on nonlinear S-form characteristics between arterial transmural pressures in aortic arch or carotid sinus and their summary baroreceptor activity. These characteristics take into consideration all known peculiarities of distinctions between threshold pressures and activity of baroreceptors from both zones:

$$\begin{aligned}
N_q^R &= \frac{1 - \exp\left(\mathbf{b}_q^R \cdot \left(\mathbf{p}_q^R - \mathbf{p}_q^t\right)\right)}{1 + \mathbf{B}_q \cdot \exp\left(\mathbf{p}_q^R \cdot \left(\mathbf{p}_q^R - \mathbf{p}_q^t\right)\right)} \\
\frac{d \Delta X_i(t)}{dt} &= \frac{K_i(t) \cdot E(t) - \Delta X_i(t)}{T_i} \\
K_i(t) &= \begin{cases} K_i^p \cdot [X_i^{\max} - X_i(t)] & E(t) > 0 \\ K_i^d \cdot [X_i(t) - X_i^{\min}] & E(t) \leq 0 \end{cases} \\
X_i(t) &= X_{oi} + M_{ij}^1 \cdot \Delta X_i(t) \\
E_q &= P_q - p_q \\
N_q^S &= a \cdot N_A^R + c \cdot N_C^R + b \cdot N_{BA}^R, a + c = 1 \\
X_i(t) &= X B_i^{\min} + \sum_{i=1}^n \Delta X_i(t)
\end{aligned} \tag{1.9}$$

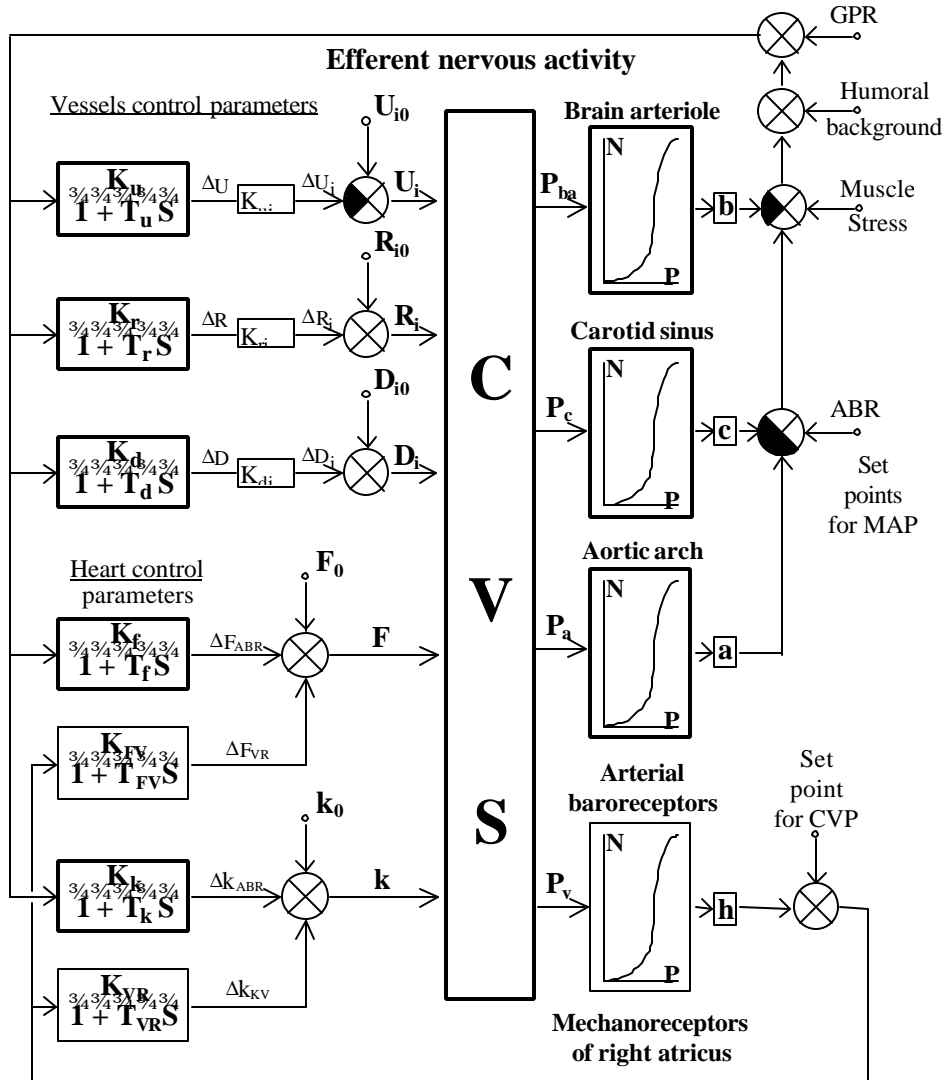


Fig.2. The general structure of model of central nervous control of CVS.

Every central reflector control mechanism was simulated as a proportional regulator that has its gain (K_i) and time constant (T_i). K_i characterizes the power, whereas T_i characterizes the value of inertia of reflector processes. We can simulate various conceivable hemodynamic situations by different combinations of changes in these parameters. It is necessary to note that every K_i does not have a constant value. There are some functional relations between K_i and control parameters (X_i): K_i decreases simultaneously with the increasing of parameters. Consequently, the functional reserves of CVS's control parameters have to decrease.

1.5 Mathematical model of the additional reflexes involving under high sustained Gz accelerations.

According to our modeling concept, we assume that there are three factors able to modify the arterial pressure's set-point level in the scheme of Fig.2. They are:

1. The concentration of different humoral cardio- and vasomotor active substances (especially catecholamines) – Y_{cat} ;
2. The general level of the body muscle activity – Y_m ;
3. The mentioned above general pressor reaction – Y_{gpr} .

The appropriate calculating formulas are looking as follows:

$$Y_{cat} = Y_{cat}^{max} \times \exp[-\mathbf{w} \times (T - T_{exp})] / \{1 + \mathbf{f} \times \exp[-\mathbf{w} \times (T - T_{exp})]\} \quad (1.10)$$

$$Y_{gpr} = K_{gpr} \times Y \times \{1 - \exp[\mathbf{h} \times (\mathbf{w}_{gpr} - A)]\} / \{1 + \mathbf{J} \times [1 - \exp[\mathbf{h} \times (\mathbf{w}_{gpr} - A)]]\} \quad (1.11)$$

$$Y_m = \mathbf{q} \cdot P_m(t) \quad (1.12)$$

The next formula presents in the ‘Pilacel’ models the dependence between the value of the emotional stress S and acceleration. It includes both male/female differences and production / utilization differences depending on acceleration changes.

$$S = \begin{cases} \mathbf{g}_m^s (1 - e^{-r_1}) / (1 + \mathbf{w} \cdot e^{-r_1}), \Delta g > 0 \\ 3 \cdot \mathbf{g}_m^s (1 + \mathbf{h} \cdot (j - 1) \cdot e^{-r_1(t' - t)}) / (1 + \mathbf{w}_1 \cdot e^{-r_1(t' - t)}), \Delta g \leq 0 \end{cases} \quad \mathbf{g}_m^s = \overline{1, 3} \quad (1.13)$$

This factor is modifying the coefficients of heart rate's baroreflexor control

$$\Delta F^S = K_g^F \cdot [1 + k_g^F \cdot (j - 1)] \\ F_N = F_c \cdot [1 + k_j^F \cdot (j - 1) + \Delta F_s \cdot (1 + k_j^a \cdot a / a_0) + K_B^F (F^{max} - F) \cdot C1 / (F^{max} - F_{min}) + \Delta F_B^B] \quad (1.14)$$

It also will modify the general output of these control mechanisms as:

$$E_B = (1 + a_s \cdot S) \cdot Y_G^{PR} - K_a^B \cdot N_a - K_c^B \cdot N_c - K_B^B \cdot N_B + K^m \cdot Y^m \quad (1.15)$$

Three next formulas were chosen to additionally approximate changes in central nervous regulators causing by age (a), sex (j) and health (h) factors.

$$E^I = N^R \cdot E_B \cdot (1 + \mathbf{a}^E \cdot a / a_0) \cdot (1 + \mathbf{f}^E (j - 1)) \cdot 1 / [1 + K^h \cdot (h - 1)] \quad (1.16)$$

$$T^H = T_B^H \cdot (1 + K_h^H \cdot (h - 1)) \cdot (1 - \mathbf{f}^H \cdot (j - 1)) \quad (1.17)$$

$$T^V = T_B^V \cdot (1 + K_h^V \cdot (h - 1)) \cdot (1 - \mathbf{f}^V \cdot (j - 1)) \quad (1.18)$$

The actual formulas for calculating of vascular tonus characteristics (resistance r , rigidity d and unstressed volume U) are followings:

$$r_n = r_n^B + K_n^r \cdot C^r \cdot C_0^V \quad (1.19)$$

$$d_n = d_n^B \cdot (1 + K_n^d \cdot C^d \cdot C_0^V) \quad (1.20)$$

$$U_n = U_n^B \cdot (1 - K_n^U \cdot C^U \cdot C_0^V) \quad (1.21)$$

where

$$C_0^V = \frac{V_{ES} \cdot [1 - \mathbf{x} \cdot (j-1)] \cdot (1+S)}{1 + \mathbf{x}^h \cdot (h-1)} \quad (1.22)$$

$$T \frac{dV_{ES}}{dt} = K_{ES} \cdot E^I - V_{ES} \quad (1.23)$$

In the last expression S reflects emotional stress level, V is the activity of efferent sympathetic nerves, E – is the output error in central contour of baroreflex.

1.6 The technology of models' individualization

To create an individual hemodynamic model, one should possess all the personal data that were used during the basic model development. In fact, this conceptual correct demand cannot be provided by the existing physiological methodology. At the same time, using general theoretical knowledge concerning average regularities, the actual medical treatment technologies create individual control of patients' health. Certainly, these empiric approaches are far from the theoretically correct solutions of the problem. But the lack of optimal approaches compels researchers and medics to use such technologies. This introduction is also applicable to the material, described in this paragraph.

So, our approach to the problem, how to create personal models, is not strongly correct. At the same time, during this development we tried to take into account mostly such regularities, which relatively are not disputable. We should note, that the used individualization technology may be considered as “draft”, and as the first iteration in this direction. There was planned to create another –“deep” individualization technology too. Some information about the second approach was presented in our interim reports. This technology had to be built on a evaluation of the person's sensibility to postural tests. It was assumed that the changes of the arterial pressure and heart rate under human position change from the horizontal to the seat position reflects the nervous control mechanism's sensitivity to the gravitational loading. This assumption has rather a solid reason. But during the further developing of the problem it became clear that this approach requires principle changes in modeling technology. To be correct, we have to set fixed changes of HR and MAP for every person, then recalculate all vessels' tonus parameters dispersed in vascular net. Such approach is acceptable for the static state, but is not available within the concept of dynamic models, used here. The acceptable model creation requires much more physiologic data, time and intellectual recourses. So, we were compelled to refuse our plan in this direction.

Conceptually, the technology of models' individualization is mainly based on the recalculation of the basic model constants, taking into account person's sex, age, weight, body length and also health. Some of these factors were included in the formulas already described above.

The main systemic variable that depends on the persons' weight and sex, is the summary blood volume V^s .

$$V_{ij}^S = K_{mij}^V \cdot (W - w_{ij}), \quad i = \overline{1,5}; j = \overline{1,2} \quad (1.24)$$

where W – is the value of personal weight in kg, K – an approximation coefficient that reflects the weight-volume relationships. Ten different values of approximation constant w were chosen depending on measured W and sex. The five values of parameter w are for the male ($j=1$) persons, and the other five values- for the female ($j=2$) persons). Thanks to such approach, models allow to calculate the individual summary blood volume V_j^S . As far as V_j^S is a starting variable of the whole CVS, it has to be distributed in each of $n = 37$ model compartments, using specially determined coefficients k_{nj}^V to the calculated value of V_j^S .

$$V_j^S = \sum_{n=1}^{37} V_{nj}, \quad j = \overline{1,2} \quad (1.25)$$

$$V_{nj} = k_{nj}^V \cdot V_j^S \quad (1.26)$$

The last variables are the basis to determine unstressed volumes of vessel compartments taking into account their sensibility to the persons' age and sex

$$U_{nj} = k_{nj}^U \cdot V_{nj} \cdot (1 + \mathbf{a}^U \cdot a / a_0) \quad (1.27)$$

To determine the values of hydraulic resistance for every n compartment in individual model, their values for the basic model were used. The recounting formula looks like:

$$r_{nj}^0 = r_n^b \cdot (1 - \mathbf{a}^r \cdot a / a_0) \cdot [1 + \mathbf{b}^r \cdot (j-1)], \quad (1.28)$$

where coefficients \mathbf{a}^r and \mathbf{b}^r are reflecting age and sex variability of vascular diameter.

It was assumed that the value of the personal volume rigidity of vessel compartment may be determined using its value in the basic model and the value of the sensitivity coefficient to the person age as:

$$d_{nj}^0 = d_n^b \cdot (1 + \mathbf{a}^d \cdot a / a_0) \quad (1.29)$$

One of the main factors, that may have essential role in determining of male – female differences in Gz tolerance, is the maximal level of muscle stress pressure P_m . This parameter also depends on the power of muscle. Usually it correlates with the person's weight, but it may be modified by a physical health, indicated in the model through the parameter h ($h = 1$ – healthy person, $h = 2$ – weak person). Taking into account these reasons, the individualization formula for P_m was created as:

$$P_j^{max} = P_b^{max} \cdot [1 + \mathbf{b}^p \cdot (j-1)] \cdot [1 - \mathbf{b}^h \cdot (h-1)] \cdot \left[1 + \mathbf{b}^w \cdot (j-1) \cdot \frac{W - w^b}{w^b} \right] \quad (1.30)$$

So, the equation system (1.1) – (1.30) presents the main mathematical relationships of the actual PM. To imagine the real complicity of the actual PM model, and problems that have been solved to reach acceptable simulation results, one should know that in fact, these equations mainly describe physiological regularities for one model compartment only. Because of the PM consists of 37 compartments, the real equation system of the PM already consists of more than 1100 basal equations. Several equations and formulas have their variants for concrete compartment too. In addition, the HPFM, numbered as (1.1), consists of 11 formulas. They are relating to the four heart's chamber. So, the HPFM is presented by 44 formulas. By analogy, the formula numbered (1.9) cover seven equations, that all together describe common baroreflexor regularities for all CVS's parameters. The EM, presented in the next chapter, essentially increases the equations quantity. Besides, there are additional logic conditions that also have to be considered as inside model information.

2 MATHEMATICAL MODELS OF THE ENVIRONMENT

According to the actual theoretical concept, all factors that may be used as protective under piloting Gz accelerations, are able to increase the top limit of human tolerance by means of such changes in human hemodynamics that result an increasing of the systemic arterial pressure. So, these factors create an additional outside artificial environment for the human cardiovascular system. This environment creates and provides extravascular forces that are able to redistribute blood volumes from lower body area towards central and head areas. Therefore, as it was marked above, it is convenient to call all these outside protective tools and methods as environmental. This chapter presents mathematical models of the artificial environment in the “PILACCEL”.

2.1 Mathematical modeling of acceleration loading.

Acceleration is the factor the changes of which are able to origin gravitational inertial forces. According to the equations numbered (1.4), gravitational forces are one of the factors that create hydrodynamic forces and initiate blood flow between two vessel compartments, connected through hydraulic resistance. The power of the gravitational force depends on the difference between levels of these vessel compartments, and is proportional to the acceleration vector's projection on the axis of blood flows.

Our model mainly simulates human person in his/her seat position. But it is also able to simulate both horizontal and erect human positions. In acting variant of the “PILACCEL” model the acceleration loading is available after a previous hemodynamic static regime for the human seat position is confirmed. To load acceleration, one should at least know the time point (t_s) to start the loading, the gradient of the acceleration increasing (A_g). He/she has also to know the top level (A_m) of the acceleration, their duration (T), and at last, the deceleration gradient (D_g). Usually, the acceleration loading profiles are constructed from the linear pieces. So, to describe acceleration profile in the model, it is enough to construct three elementary pieces: increase, plateau and decrease.

The first elementary profile is presented in the “PILACCEL” as:

$$a = \begin{cases} A_g \cdot (t - t_s), & t > t_s \\ 0, & t \leq t_s \end{cases} \quad (2.1)$$

The plateau part is described as:

$$a = A_m, \quad t_p < t < t_p + T, \quad (2.2)$$

where $t_p = A_m / A_g$.

By analogy with the first piece, the third piece describes linear decreasing of accelerations as:

$$a = \begin{cases} B_g \cdot (t - t_R), & t_p + T \leq t \leq t_R, \quad t_R = A_m / B_g \\ 0, & t > t_R \end{cases} \quad (2.3)$$

But the real dynamics of the acceleration changes has some specific differences comparing with the described linear characteristics. There are two factors causing these differences. At first, experts know that because of inertia neither centrifuge nor flight is able to change its running parameters strongly linear. Usually, they have some nonlinear forms both on the start part and close to the stabilization of the plateau acceleration level. The second factor, that will be included in the model to be able simulate the nonlinear behavior of real accelerations, is caused by dimensional change of human position relatively to the Earth

gravitational vector. Really, the gravitational loading value $G = 1$ acts on the human person while he/she is sitting on the armchair in the centrifuge before to be loaded by acceleration. This position is typical also for the pilot under his/her horizontal flight. But the first few instants of acceleration loading this position changes so, that the Earth gravity direction is not more similar to the direction of the acceleration vector. Certainly, this is a peculiarity only, but it has its influence on the acceleration loading dynamics.

The first mentioned above factor was included in the model by means of differential equation:

$$T^{in} \cdot da/dt = A_g \cdot t - a, \quad (2.4)$$

where T^{in} characterizes the running inertia.

The effects of the second factor is included in the model using a special notion of bank angle

$$\Omega = \begin{cases} 0.5 \mathbf{p} \cdot \mathbf{e}^{-a}, a > 0 \\ 0, a \leq 0 \end{cases} \quad (2.5)$$

The acting gravitational component of the hydrodynamic forces is calculated using this formulae for real acceleration determining:

$$Ra = 1 + a - \cos \Omega \quad (2.6)$$

2.2 Modeling of the armchair's protective function.

Theoretically, the armchair has three independent parameters that may be used to influence on the top limit of the human tolerance to G_z accelerations. These parameters are the angles between G_z vector and seat-back angle, thigh angle, shank angle. To be able to change these angles in the model, we have specially presented the thigh and the shank vessel compartments (see Fig. 1) in vascular net model. It is assumed that the head and the supine have common direction. So, this model excludes independent changes of human head relatively to torso. It is known, pilots sometimes use head-down shifting to resist G_z acceleration. Perhaps, this limit may play some role during analysis of factors determining different values of simulation and real human tolerance to $+G_z$ acceleration.

2.3 Mathematical modeling of the anti-G suits.

In frame of this development we have had to create models of two principally different anti-G suits. The first model will describe the protective effects of the usual pneumatic trousers, while the second model will describe the protective effects of new 'Libelle' anti-G suit. The last suit contains special liquid volumes and acts without air inflation by automatic increasing of local hydrostatic pressures under gravitational and inertial forces.

2.3.1 Pneumatic anti-G suit.

In the model of main hemodynamic effects of the use anti-G-trousers the pressure transmission process from sections (three sections maximum are available) of trousers into the tissue around vessels is presented with the help of the following equations:

$$\begin{aligned} T \cdot \frac{dP^I(t)}{dt} &= K \left(P^I(t) \right) \cdot P^E(t) - P^I(t), \quad g \geq g_T^{th} \\ P^I(t) &= 0, \quad g < g_T^{th} \end{aligned} \quad (2.7)$$

where T -time constant and K- transmission coefficient of different human body cavities or tissues, g_T^{th} -level of threshold of acceleration +Gz to start trousers' inflation. PE and PI are air pressure in sections and extravascular pressure in tissue (or cavity) respectively. The user can set different values of coefficients K for every suit's section, simulating different constructions of pneumatic anti-G suits.

2.3.2 “Libelle” anti-G suit.

Although the developers are asserting that the “Libelle” anti-G suit practically has not an activation inertia, we cannot agree with them. Certainly, this inertia may be essentially less than in pneumatic suits (where it mainly causes by compressor delays), but the “Libelle” anti-G suit has its own inertia too. This inertia is caused by space between human body and suit. This space may be minimized, but it could not be absolutely excluded. Such kind of data, necessary to create an adequate model, is absent. Therefore, we assumed that the dynamics of inside P_{lib} pressure changes in the “Libelle” under changes of accelerations may have usual S-form saturation curve and may be presented as:

$$P_{\text{lib}} = \frac{1 - e^{-bRa}}{1 + (2 - R^f) \cdot e^{-bRa}} \quad (2.8)$$

$$P_{\text{lib}}^a = \mathbf{h} \cdot H_{\text{ab}}^a \cdot P_{\text{lib}} \cdot S_{\text{aa}} \quad (2.9)$$

$$P_{\text{lib}}^h = H^h \cdot P_{\text{lib}} \cdot S_{\text{aa}} \quad (2.10)$$

$$P_{\text{lib}}^T = P_{\text{lib}}^a + \mathbf{h} \cdot H_t^a \cdot P_{\text{lib}} \cdot S_{\text{aa}} \quad (2.11)$$

$$P_{\text{lib}}^{\text{Sh}} = P_{\text{lib}}^T + \mathbf{h} \cdot H^{\text{Sh}} \cdot P_{\text{lib}} \cdot S_s \quad (2.12)$$

$$S_{\text{aa}} = R_a \cdot \cos A; \quad S_T = R_a \cdot \cos B; \quad S_s = R_a \cdot \cos C \quad (2.13)$$

Where A, B, C are angles between acceleration vector and body torso, thigh and shank parts respectively. H^{Sh} , H_t^a , H_{ab}^a , H^h are lengths of shank, thigh, abdominal and hand vessel compartments respectively.

Depending on the value of coefficient \mathbf{h} , that reflects the top level of liquid in “Libelle” suit body section, there may occur different protective situations. These situations were special analyzed and described in paragraph 4.1.

2.4 Mathematical modeling of the breathing regimes.

Special breathing regimes are proposed to increase human tolerance to Gz acceleration [5, 7]. The positive pressure breathing (PPB) is one of the main standard ways. Usually, such protection method's using is effective in combination with the use of anti-G suits. This method may be used by the person, who is able to control his/her body muscle state in its relaxed or peripheral stressed condition.

Before describing the model of positive pressure, we will note that our general model includes also specific dependence of the pleural pressure from the +Gz accelerations as:

$$Pp(t) = Pp_0 - I \cdot g(t) \quad (2.14)$$

The empiric investigations show that it is worth when this method is added after the acceleration level is about four units. The recommended profile for the breathing pressure linearly increases until 60 mm Hg plateau. The gradient of the breathing pressure increasing and the threshold of the breathing pressure to start its increasing may be changeable. Our model presents this protective method by means of formula that determines the pleural pressure Pp linearly increasing starting at some +Gz-threshold level:

$$P_p(t) = \begin{cases} P_{p_0}, & g \leq g_{BP}^{th}, \\ P_{p_0} + a_{BP} \cdot \left(g - g_{BP}^{th}\right), & g > g_{BP}^{th} \text{ \& } P_p(t) \leq P_{BP}^{max}, \\ P_{BP}^{max}, & P_p(t) > P_{BP}^{max} \end{cases} \quad (2.15)$$

The actual model also includes two additional opportunities that were absent in previous development. Our basic models were developed in assumption that the pleural pressure has not any breathing pulsation. To be able simulate hemodynamic effects of the breath pulsation, our actual model presents a sinusoidal time changes for the pleural pressure. Because of the amplitude (A_b) and the frequent (f_b) of these pulsation may be voluntary changed by human person, these parameters may be set by user. In versus time, they are fixed

$$Pp(t) = Pp_0 + \mathbf{s} \cdot A_b \sin(\mathbf{p} \cdot f_b \cdot F \cdot t) \quad (2.16)$$

Coefficient sigma allows to describe the main effects of $Pp(t)$ modulations by the heart rate (F).

The second additional opportunity mentioned above is the ability to simulate every thinkable breathing regimes. The last formula provides a simple way for such ability (one can set parameters A_b and f_b and compel model to “breath” how he/she wants). Another way is to create an arbitrary breathing profile using special option of the user interface (see later).

2.5 Mathematical modeling of the body muscle stress.

In frame of this development we are interested in body muscle stress's hemodynamic effects only. They may occur under changes of the local extravascular pressures. This factor was included in our model of the vascular hemodynamics as one of the three independent forces that form local integral blood pressure. So, the local muscle stress would origin changes of the local blood pressure and thus could change blood flow. As far as our models are dynamic, we have to describe the dynamics of muscle stressing, determining the starting and final pressure levels and also the form of this transforming. The problem is that these characteristics are strongly individual. Besides, in the investigations relating to the problem of human gravitational tolerance, such characteristics never have been measured. Therefore, the only way to model our problem is use of euristics.

The formula used in our early model was created in assumption that the mean muscle tense (P_m) will linearly increase starting from some acceleration threshold. It looks as following:

$$P_m = \begin{cases} 0, & g \leq g_m^{th} \\ A \cdot P_m^{max} \cdot \left(g - g_m^{th}\right) g_m^{th} < g \leq P^{max} \\ P_m^{max}, & g > P^{max} \end{cases} \quad (2.17)$$

$$Pp(t) = Pp_0 + \mathbf{s} \cdot A_b \sin(\mathbf{p} \cdot f_b \cdot F \cdot t)$$

To describe the nonlinear S-form dynamic effects of muscle pressure $P_m(t)$ increasing under muscle stress the new formula was found out:

$$P_m(t) = P_m^{max} \cdot [1 - \mathbf{x}_p^h \cdot (h - 1)] \cdot [1 - \mathbf{x}_j^p \cdot (j - 1)] \cdot \frac{1 - e^{-\mathbf{q}^p [q_m^p - G(t)]}}{1 + \mathbf{h}^p \cdot e^{-\mathbf{q}^p [q_m^p - G(t)]}} \quad (2.18)$$

It describes the possible changes in the dynamics of P_m for the male ($j = 1$) or female ($j = 2$) persons, and also for the healthy ($h = 1$) or weak ($h = 2$) persons.

The block of muscle stress models consists of four different models. The first model describes the changes of the regional $P_m(t)$ in the hand, abdomen area under special protection called a hand gripping. It also describes the pleural pressure changes under this procedure.

The second model describes the special situation that is known as peripheral stressing. We think that this state occurs rather often under real flight with the moderate level of accelerations. To model this situation, we suppose that the level of reached pressure is about $0.85 P_m^{\max}$ in (2.18).

The third model describes the AGSM. It consists of three equations. The first one is similar to (2.18) but presents pleural pressure increasing under muscle stress. The second equation describes the pleural pressure decreasing under rapid inspiration, while the third equation describes the dynamics of the pleural pressure's rapid recovering to the strained level.

$$P_p(t) = \begin{cases} P_{p_0} + P_m \cdot \frac{1 - e^{d(t-t_s)}}{1 + z \cdot e^{d(t-t_s)}}, & t > t_s \\ P_{p_0} - P_m \cdot \frac{1 - e^{q(t-t_i)}}{1 + z_2 \cdot e^{q(t-t_i)}}, & t_s \leq t \leq t_i, \\ P_{p_0} + (P_m + 1) \cdot \frac{1 - e^{q(t-t_e)}}{1 + z_2 \cdot e^{q(t-t_e)}}, & t > t_i, \end{cases} \quad (2.19)$$

2.6 Modeling technology of human protection under Gz accelerations.

All four main methods of human artificial protection under high sustained +Gz accelerations (changes of angles between human body parts and direction of acceleration vector, use of anti-G suits, muscle tension, special breathing regimes under positive pressure), described above, were included in our final software model complex. It is important to note that they are additive. The future user will be able to choose each of the possible combinations of protections from this list. There are several combinations that cannot be realized. These situations are specially marked and the user will see warnings every time trying to simulate such combinations.

To simulate a simple computer experiment, one should choose the model type (basic or individual) and to set all input data.

3 THE MAIN TEST RESULTS

The “PILACCEL” presents to user an ability to do a simple computer simulation experiment, changing at least one from the 44 model characteristics. So, the quantity of the potentially possible variants is a giant number, and nobody could be able to realize all possible combinations! At the same time, to be sure that simulation results are reasonable, several test situations are presented here.

Because of models and test loading are dynamic, one of the main problems was the test regime choice. Let's try to explain origin of this problem.

The “PILACCEL” provides simulation scenario similar to empiric research technology, developed to determine human tolerance to +Gz acceleration, creating by means of centrifuge. So, it is possible to simulate every thinkable acceleration profile. The critical event, which causes emergence change of acceleration loading both in simulation and nature experiments, is central light loss (CLL). During the empiric observations the systolic arterial pressure (PES) in ear level is controlled to predict the CLL. Experts agreed that the critical low limit of the PES is about 30 mm Hg. But usually, the eye retinal cells and brain cells may continue their function about 1-2 sec after the blood flow is blocked. When one wants to obtain the human tolerance to gradual increasing acceleration (for example, the increasing gradients are 0,1 – 0,2 g/s), these 1-2 sec are not essential. But principally another situation will occur when we use this evaluation method under the rapid acceleration with the gradients, more than 1 - 4 g/s. The evaluation error may sometimes reach up to 8 G, if we use the PES under linear profile acceleration loading with gradient 4g/s. So, we have to differ two definition for the acceleration tolerance: one – for the short time tolerance (< 2 sec), and another – for the long time (sustained) tolerance (> 2 sec). Under the rapid linear increasing acceleration the human short time tolerance is essentially higher, than for the stable acceleration level, but with the duration longer than 2 sec. The trapezoid profile will be considered as the most convenient acceleration's loading regime, able to display the human tolerance to the sustained accelerations. So, the results, presented and discussed below, are related to the trapezoid acceleration profile.

To demonstrate the adequacy of the “PILACCEL” models, following main regimes were chosen by us and presented here:

- relaxed mean man without protections but for three different values of seat back angle;
- relaxed mean man with pneumatic anti-G suit protection;
- relaxed mean man with positive pressure breathing (PPB) & pneumatic anti-G suit protection;
- peripheral muscle stressed mean man with pneumatic anti-G suit protection;
- antigravitational stressed maneuver (AGSM) & pneumatic anti-G suit protection;
- individual male model without and with the pneumatic anti-G suit protection;
- individual female model without and with the pneumatic anti-G suit protection;

For the several positions the blood volumes' redistribution also are shown. Besides, special regime, that is known as “Push-Pull” profile, also is presented. As to the ‘Libelle’ anti-G suit protective effects, they are presented and analyzed in the next chapter.

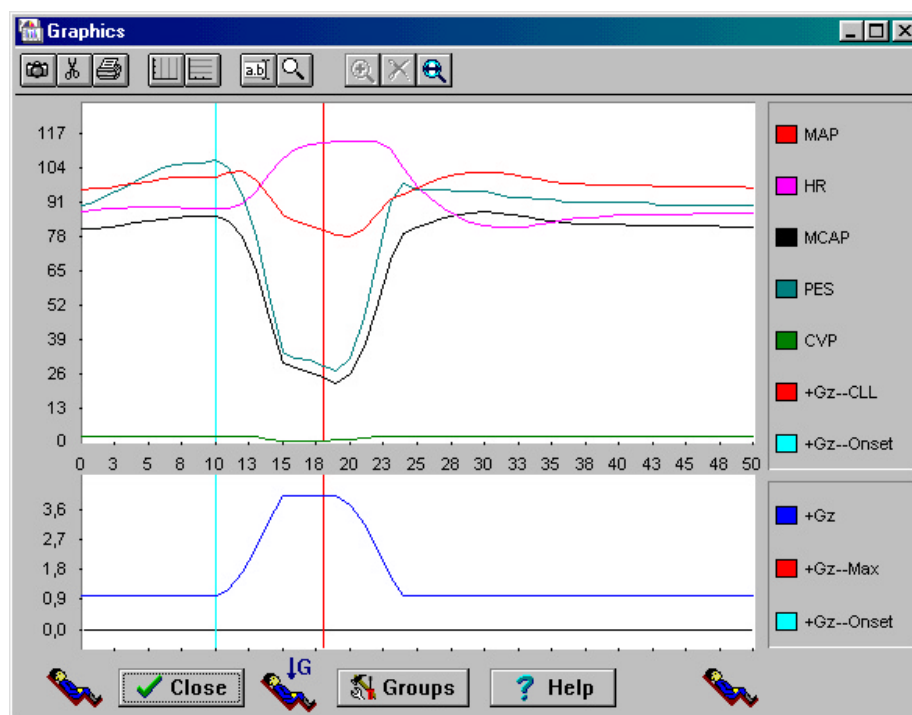


Fig.3. The dynamics of mean arterial pressure (MAP), mean pressure in carotid sinus (MCAP), systolic arterial pressure on the eye level (PES), central venous pressure (CVP) and heart rate (HR) under trapezoid profile accelerations with the acceleration and deceleration gradients 1g/s (the bottom part of the figure).

The mean relaxed man. The emotional stress regime is set on moderate position. Seat back angle = 12° .

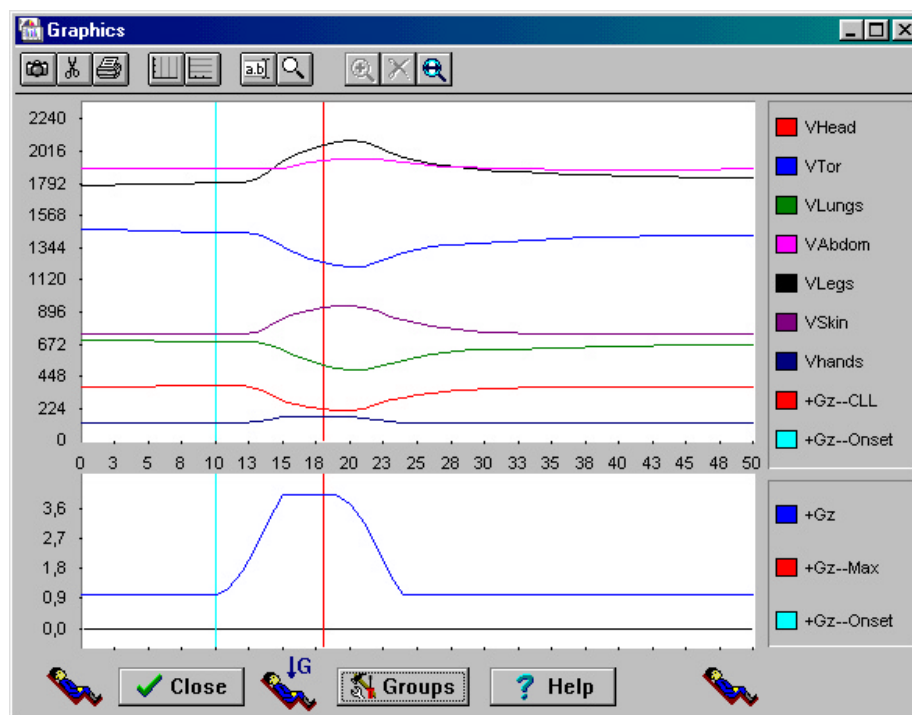


Fig. 4. The dynamics of blood volumes in different body sections (Vtor – thorax; Vabdom – abdominal; VLungs – lungs; Vlegs- legs; Vhead – head; Vhand –Hands; Vskin – Skin) under trapezoid profile accelerations with the acceleration and deceleration gradients 1g/s (the bottom part of the figure).

The mean relaxed man. The emotional stress regime is set on moderate position. Seat back angle = 12° .

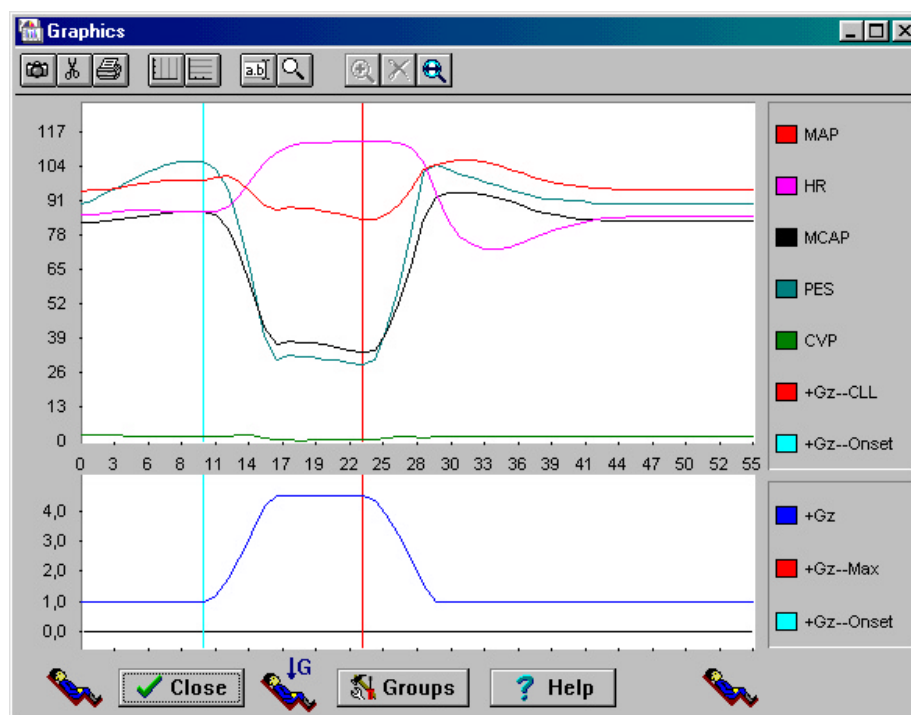


Fig.5. The dynamics of mean arterial pressure (MAP), mean pressure in carotid sinus (MCAP), systolic arterial pressure on the eye level (PES), central venous pressure (CVP) and heart rate (HR) under trapezoid profile accelerations with the acceleration and deceleration gradients 1g/s (the bottom part of the figure).

The mean relaxed man. The emotional stress regime is set on moderate position. Seat back angle = 45° .

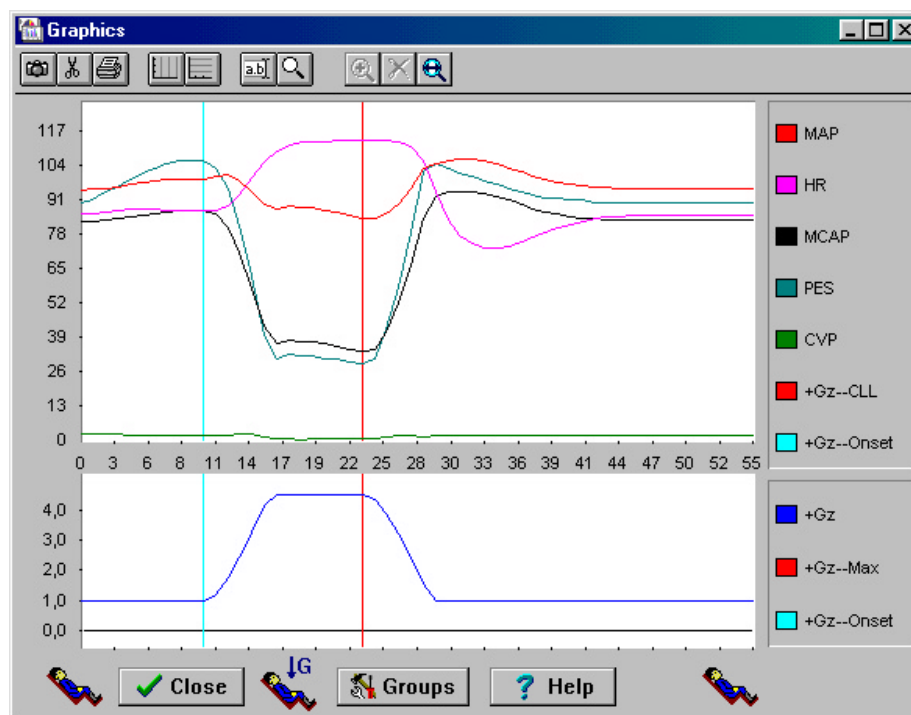


Fig.6. The dynamics of mean arterial pressure (MAP), mean pressure in carotid sinus (MCAP), systolic arterial pressure on the eye level (PES), central venous pressure (CVP) and heart rate (HR) under trapezoid profile accelerations with the acceleration and deceleration gradients 1g/s (the bottom part of the figure).

The mean relaxed man. The emotional stress regime is set on moderate position. Seat back angle = 75° .

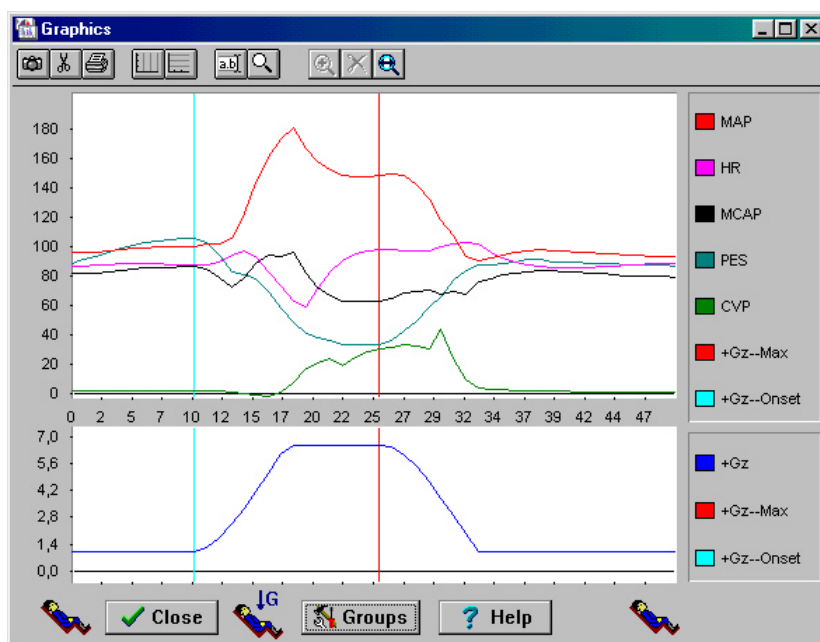


Fig.7. The dynamics of mean arterial pressure (MAP), mean pressure in carotid sinus (MCAP), systolic arterial pressure on the eye level (PES), central venous pressure (CVP) and heart rate (HR) under trapezoid profile accelerations with the acceleration and deceleration gradients 1g/s (the bottom part of the figure).

The mean relaxed man with use of pneumatic anti-G suit. The suit is pressured with gradient 1.5 Psi after the acceleration exceeds 2g. Sections' covering percents are the following: abdominal –35%; thigh –75%; shank – 90%.

The emotional stress regime is set on moderate position. Seat back angle = 30°.

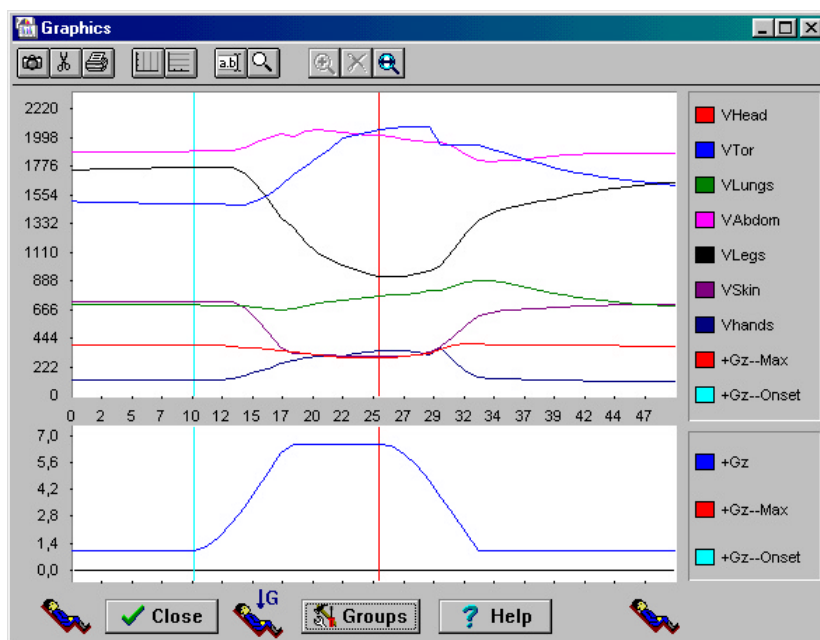


Fig.8. The dynamics of blood volumes in different body sections (Vtor – thorax; Vabdom – abdominal; Vlungs – lungs; Vlegs – legs; Vhead – head; Vhand – Hands; Vskin – Skin) under trapezoid profile accelerations with the acceleration and deceleration gradients 1g/s (the bottom part of the figure).

The mean relaxed man with use of pneumatic anti-G suit. The suit is pressured with gradient 1.5 Psi after the acceleration exceeds 2 g. Sections' covering percents are the following: abdominal –35%; thigh –75%; shank – 90%.

The emotional stress regime is set on moderate position. Seat back angle = 30° .

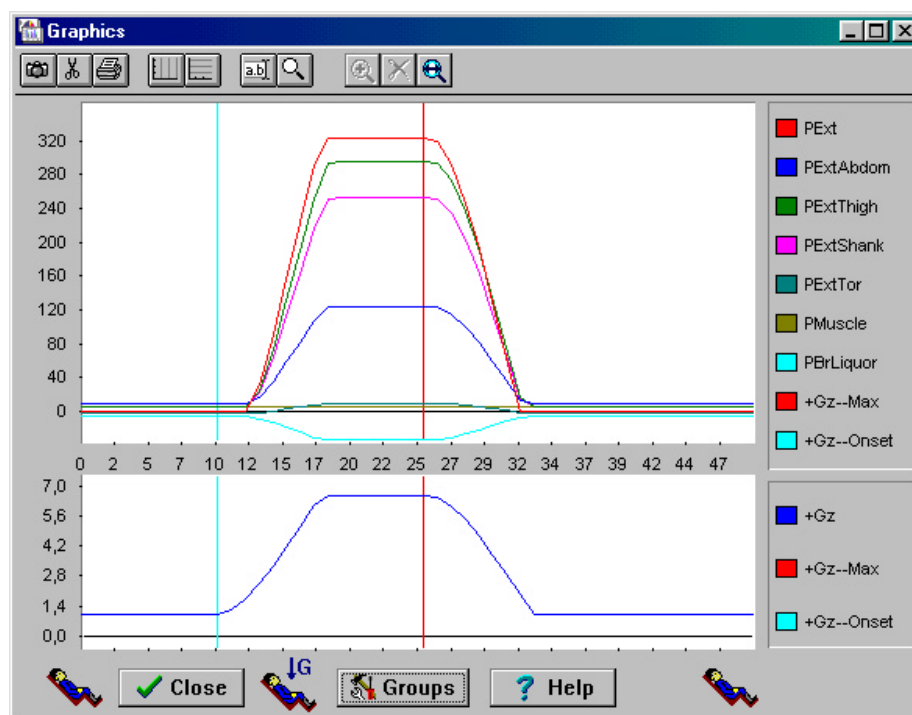


Fig.9. The dynamics of pressures, generated by compressor (Pext) and transmitted into abdominal cavity (PExtAbdom), thigh (PExtThigh) and shank (PExtShank) sections of suit, and also pleural pressure (PextTor), muscle pressure (Pmuscle), brain liquor pressure (PLiquor) under trapezoid profile accelerations with the acceleration and deceleration gradients 1g/s (the bottom part of the figure).

The mean relaxed man with use of pneumatic anti-G suit. The suit is pressured with gradient 1.5 Psi after the acceleration exceeds 2 g. Sections' covering percents are the following: abdominal –35%; thigh –75%; shank – 90%.

The emotional stress regime is set on moderate position. Seat back angle = 30° .

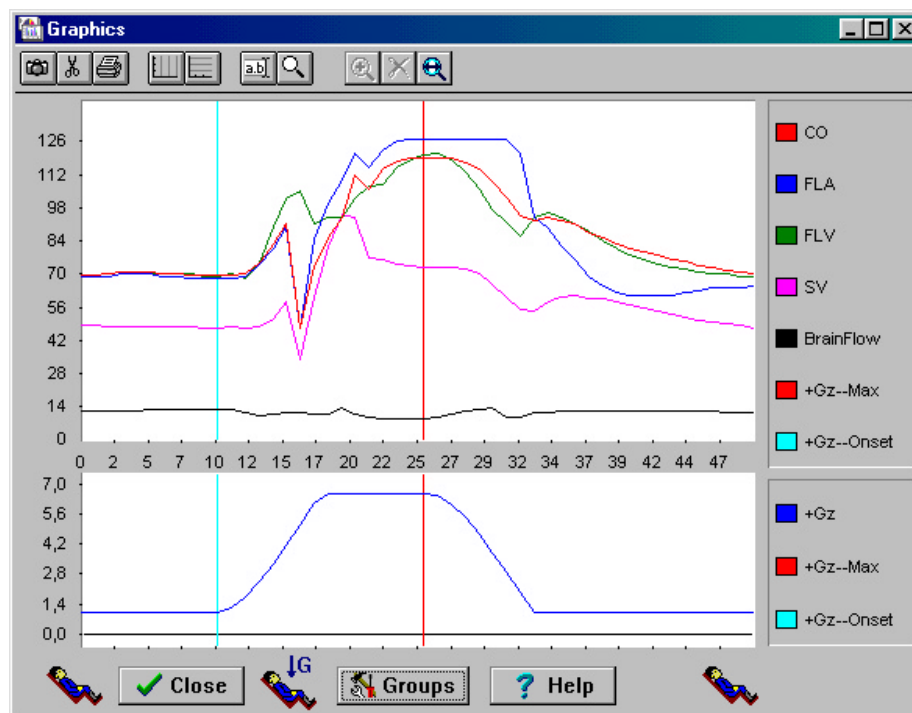


Fig.10. The dynamics of cardiac output (CO), Stroke volume (SV), lung arterial (FLA) and venous (FLV) flows, and also summary brain flow (BrainFlow) under the same conditions that have been presented on Fig.7 –Fig.9.

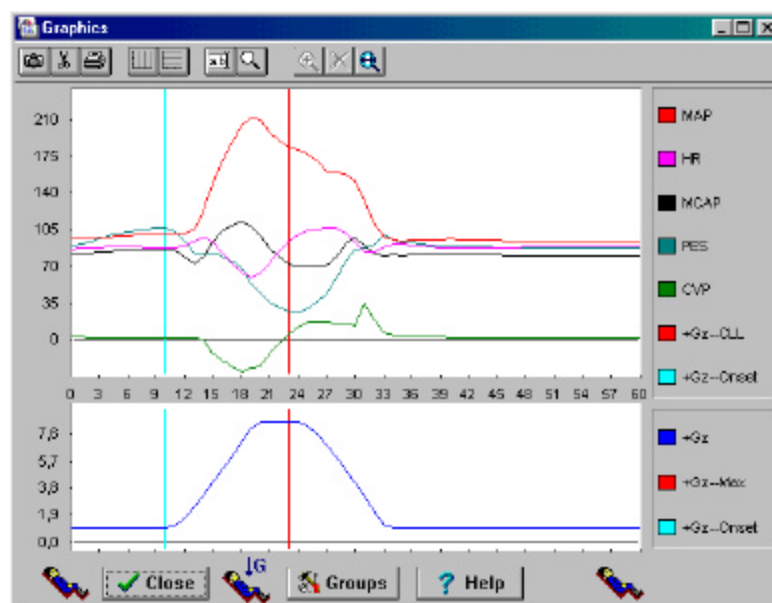


Fig.11. The dynamics of mean arterial pressure (MAP), mean pressure in carotid sinus (MCAP), systolic arterial pressure on the eye level (PES), central venous pressure (CVP) and heart rate (HR) under trapezoid profile accelerations with the acceleration and deceleration gradients 1g/s (the bottom part of the figure).

The mean relaxed man with use of pneumatic anti-G suit & PPB.

The suit is pressured with gradient 1.5 Psi after the acceleration exceeds 2 g. Sections' covering percents are the following: abdominal –35%; thigh –75%; shank – 90%.

The PPB regime is: G-threshold = 4g; Pmax = 60 mm Hg; PPB-gradient =15 mm Hg.

The emotional stress regime is set on moderate position. Seat back angle = 30° .

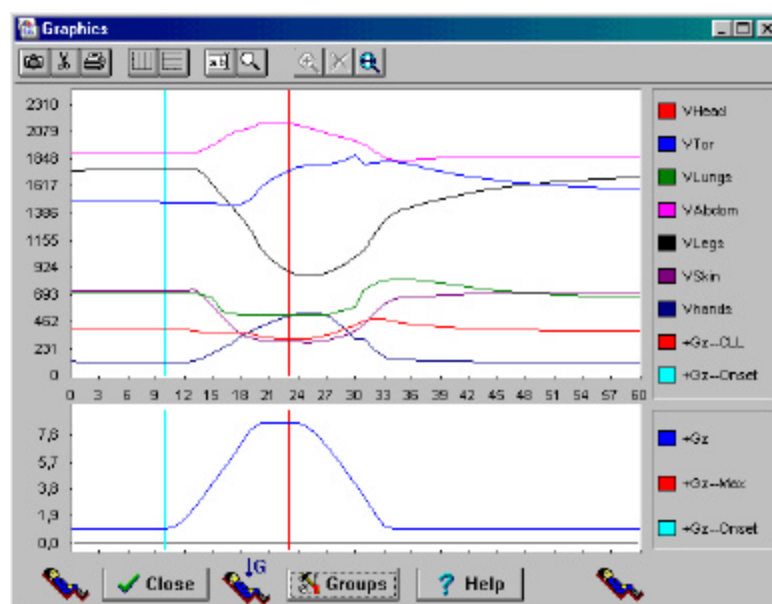


Fig.12. The dynamics of blood volumes in different body sections (Vtor – thorax; Vabdom – abdominal; Vlungs – lungs; Vlegs- legs; Vhead – head; Vhand – Hands; Vskin – Skin) under trapezoid profile accelerations with the acceleration and deceleration gradients 1g/s (the bottom part of the figure).

The mean relaxed man with use of pneumatic anti-G suit & PPB.

The suit is pressured with gradient 1.5 Psi after the acceleration exceeds 2 g. Sections' covering percents are the following: abdominal –35%; thigh –75%; shank – 90%.

The PPB regime is: G-threshold = 4g; Pmax = 60 mm Hg; PPB-gradient =15 mm Hg.

The emotional stress regime is set on moderate position. Seat back angle = 30° .

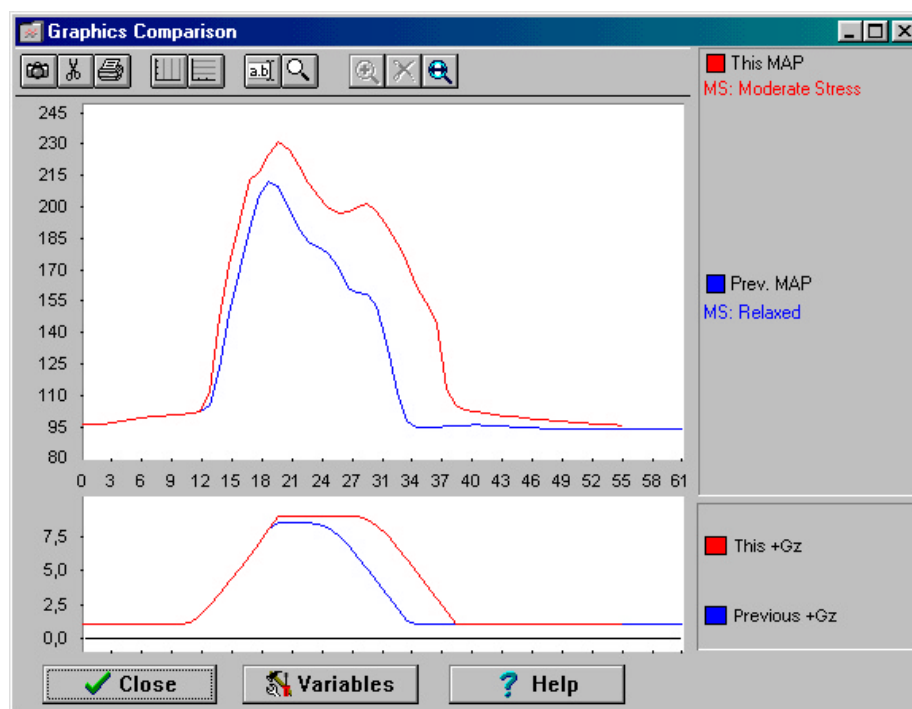


Fig.13. Comparison of MAP for muscle relaxed and muscle moderate stressed situations. All other experiment conditions are similar to case, shown on Fig.12.

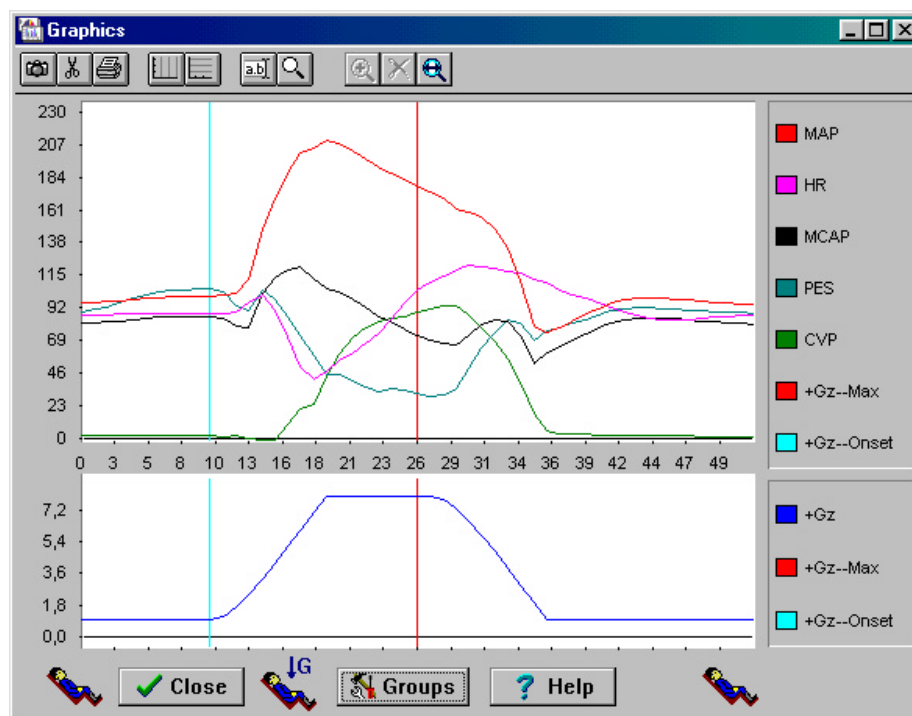


Fig.14. The dynamics of mean arterial pressure (MAP), mean pressure in carotid sinus (MCAP), systolic arterial pressure on the eye level (PES), central venous pressure (CVP) and heart rate (HR) under trapezoid profile accelerations with the acceleration and deceleration gradients 1g/s (the bottom part of the figure).

The mean man with moderate stressed muscle & use of pneumatic anti-G suit. The suit is pressured with gradient 1.5 Psi after the acceleration exceeds 2 g. Sections' covering percents are the following: abdominal –35%; thigh –75%; shank – 90%.

The emotional stress regime is set on moderate position. Seat back angle = 30° .

Comparing Fig.12 with Fig.8, one can see the differences in thoracic volumes that origin the PPB. Both the lung volume and summary thoracic volume are less than under the natural breathing. Such kind of comparison is acceptable. To do it, the user must first remember the actual experiment results. Then he/she will change some of interface parameters concerning conditions of experiment (these conditions may be related to physiology model and/or to environmental model. The next figure demonstrates such comparison for the MAP for the experiment condition that was set in previous case, and changing human muscle stress regime to its peripheral stressing state.

To see the role of muscle moderate stressing, one should compare this figure with the analogic curves shown on Fig.7. Such comparison will show that the acceleration tolerance level increases on 0,6 unit.

The next figure shows the role of AGSM maneure, added to use of standard anti-G suit.

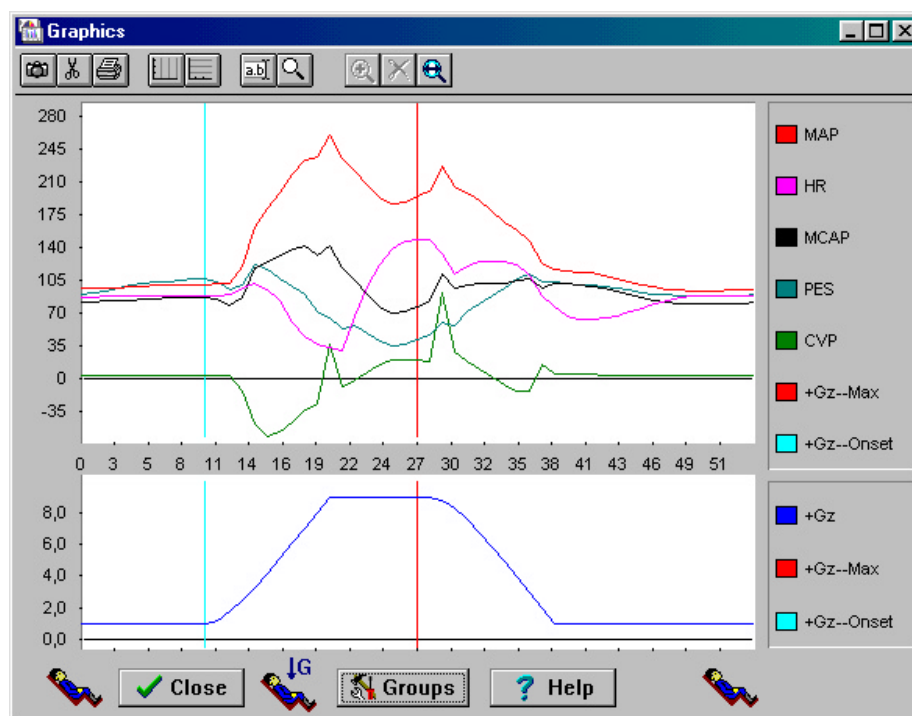


Fig.15. The dynamics of mean arterial pressure (MAP), mean pressure in carotid sinus (MCAP), systolic arterial pressure on the eye level (PES), central venous pressure (CVP) and heart rate (HR) under trapezoid profile accelerations with the acceleration and deceleration gradients 1g/s (the bottom part of the figure).

The mean man with AGSM & use of pneumatic anti-G suit. The suit is pressured with gradient 1.5 Psi after the acceleration exceeds 2 g. Sections' covering percents are the following: abdominal –35%; thigh –75%; shank – 90%.

The AGSM parameters are: Pmax = 100 mm Hg; Straining duration = 7 sec; Rapid Inspiration-Expiration duration = 2 sec.

The emotional stress regime is set on moderate position. Seat back angle = 30° .

The peaks, that one can see on CVP and MAP graphics, are caused by rapid inspiration-expiration procedure on a stable AGSM phone. These pressure changes are visible also on the curves of third group, presented on the next figure.

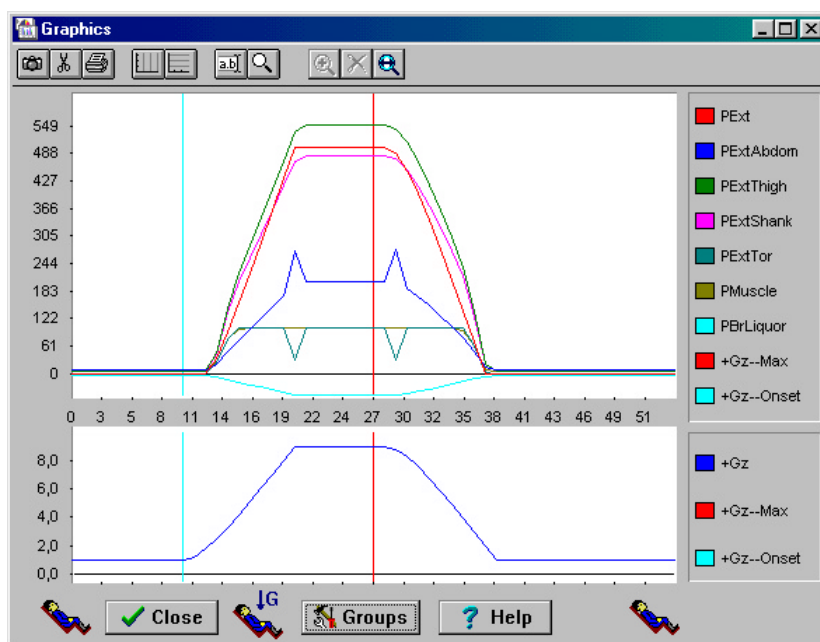


Fig.16. The visible effects of used AGSM.

Figures 3 – 16 are mainly demonstrated simulation results. Let's consider several additional cases that will show the “Push-Pull” effect, the flight with arbitrary acceleration profile. Besides, the effect of the special interface option, that transforms usual presentation to the case able to reflect breathing effects of hemodynamic characteristics, will be shown.

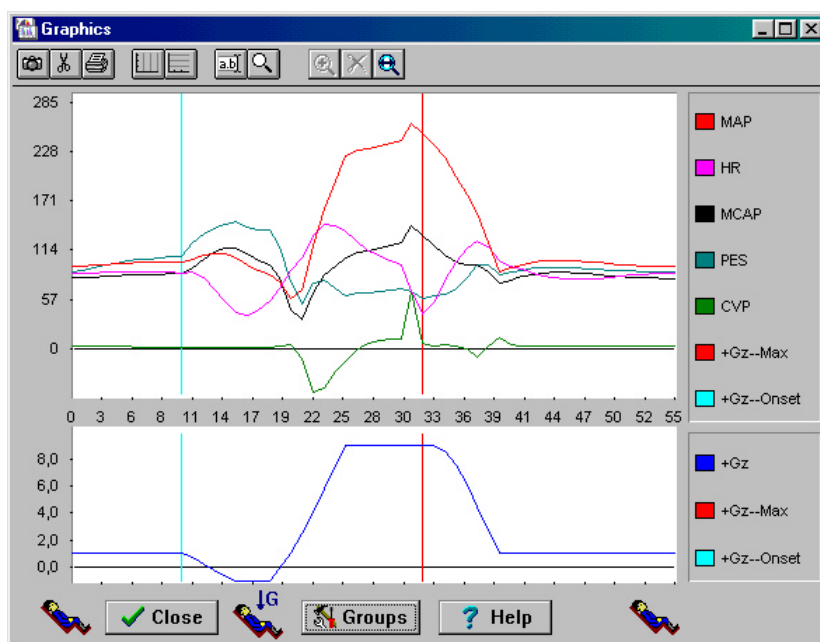


Fig.17. The dynamics of mean arterial pressure (MAP), mean pressure in carotid sinus (MCAP), systolic arterial pressure on the eye level (PES), central venous pressure (CVP) and heart rate (HR) under “Push-Pull” profile accelerations with the acceleration and deceleration gradients 2g/s (the bottom part of the figure).

The mean man with AGSM & use of pneumatic anti-G suit. The suit is pressured with gradient 1.5 Psi after the acceleration exceeds 2 g. Sections' covering percents are the following: abdominal –35%; thigh –75%; shank – 90%.

The AGSM parameters are: Pmax = 100 mm Hg; Straining duration = 10 sec; Rapid Inspiration-Expiration duration = 2 sec.

The emotional stress regime is set on moderate position. Seat back angle = 30° .

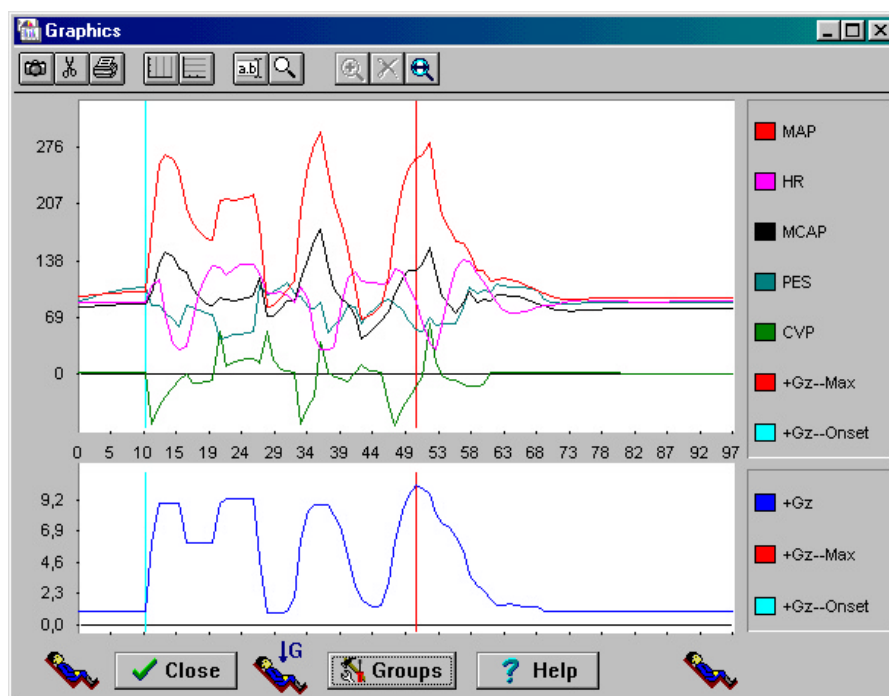


Fig.18. The dynamics of mean arterial pressure (MAP), mean pressure in carotid sinus (MCAP), systolic arterial pressure on the eye level (PES), central venous pressure (CVP) and heart rate (HR) under arbitrary profile accelerations.

All protections are identical to the case described above for Fig.17.

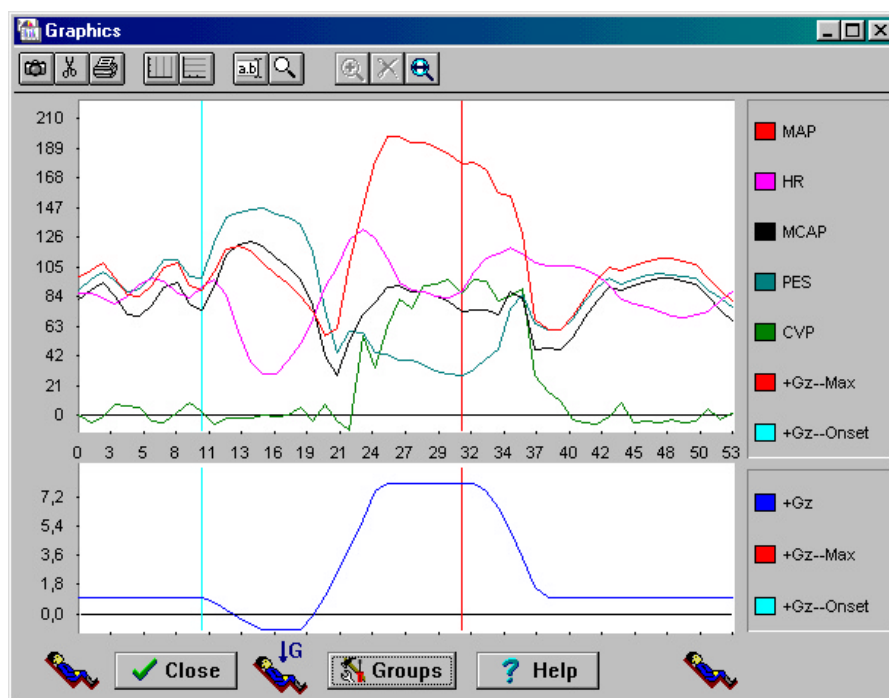


Fig.19. The dynamics of mean arterial pressure (MAP), mean pressure in carotid sinus (MCAP), systolic arterial pressure on the eye level (PES), central venous pressure (CVP) and heart rate (HR) under “Push-Pull” profile accelerations with the acceleration and deceleration gradients 2g/s (the bottom part of the figure).

The mean man with peripheral muscle stress & use of pneumatic anti-G suit. The suit is pressured with gradient 1.5 Psi after the acceleration exceeds 2 g. Sections' covering percents are the following: abdominal –35%; thigh –75%; shank – 90%.

The emotional stress regime is set on moderate position. Seat back angle = 30°.

The interface option “Show breathing pulsation” is activated.

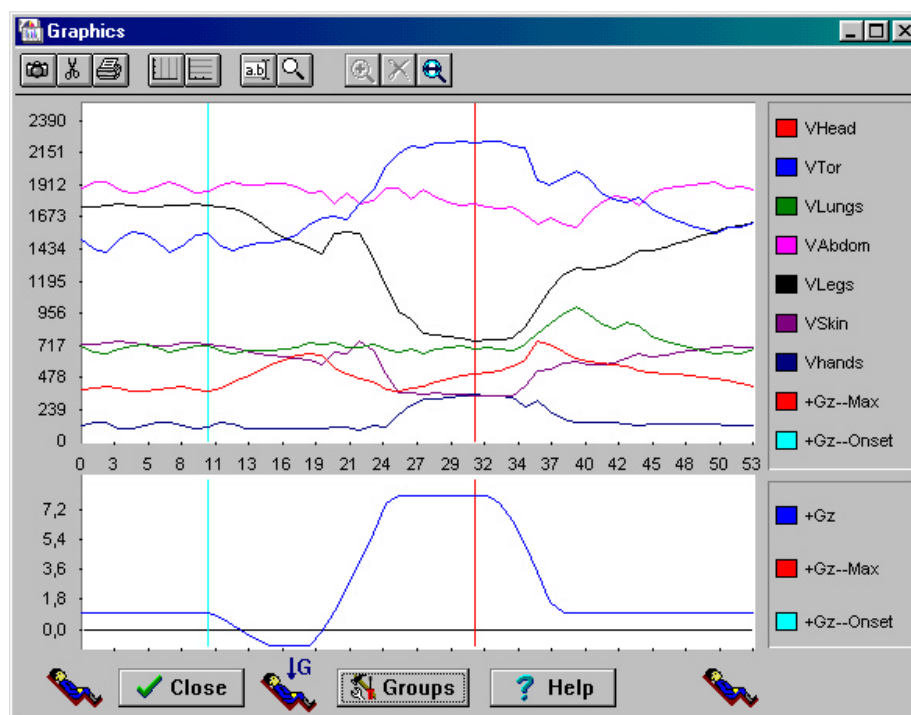


Fig.20. The dynamics of blood volumes in different body sections (Vtor – thorax; Vabdom – abdominal; VLungs – lungs; Vlegs- legs; Vhead – head; Vhand –Hands; Vskin – Skin) under under “Push-Pull” profile accelerations with the acceleration and deceleration gradients 2g/s (the bottom part of the figure).

The mean man with peripheral muscle stress & use of pneumatic anti-G suit. The suit is pressured with gradient 1.5 Psi after the acceleration exceeds 2 g. Sections’ covering percents are the following: abdominal –35%; thigh –75%; shank – 90%.

The emotional stress regime is set on moderate position. Seat back angle = 30° .

The interface option “Show breathing pulsation” is activated.

4 THEORETICAL ANALYSIS AND DISCUSSION OF SOME PROBLEMS

The model that is able to substitute the real object - prototype for the class of situations, in which investigator is interested in, may be considered as an ideal model. But this condition is relatively easy to reach for the simple models, and for very narrow class of problems only [8, 9, 18]. The “PILACCEL” is the mostly detailed model in its class. It presents to user wide range of simulation opportunities. Naturally, the question, concerning validity of these model results, will arise during its exploitation by future users. This question is actual also for us – creators. We cannot be sure that the draft individualization technology used by us here is able to provide high adequacy of simulation results for personal models. We have proposed these models as the first step, necessary to do on a way to create more adequate models. But such model creation requires special previous empiric research. Perhaps, the modeling efforts may be maximal when such research will be planned in frame of collaboration between customer and modelers. This collaboration is necessary to determine all goals of future model use and also to plan appropriate experiments able to main all basic data to develop and test the model.

Certainly, the main principles used during individual model creation are physiologically well based. But nobody in World is possessing now by data enough to be sure that all these principles could be right for the high sustained acceleration conditions. The additional empiric data, presented by customers and used by us to validate some model characteristics, were related to the HR dynamics under trapezoid profile flight. There were 4 female and 3 mail relaxed persons observed without protection use. Our actual model satisfactorily reflects these observations. But we have to not that such empiric basis is poor and far from the data necessary to proof models in all their range of acceleration and protection loading.

4.1 Theoretical analysis of the “Libelle” anti- G suit’s protective effects

To the main specific characteristics of the ‘Libelle’ anti-G suit is considerate the assumption that theoretically under its use both the inside and outside vascular hydrostatic pressures have to equally change proportionally to accelerations level [1,20,21]. This assertion is strongly right for the case only, when the hydrostatic indifferent point (HIP) in CVS and the top liquid level in torso section of the suit are equal. But this condition may not be real under suits’ use. There are several reasons to think so. The first reason concerns to dimensional stability of HIP.

The necessity and definition of the HIP was proposed and mostly used during investigations of the localization and potential role of volume receptors in CVS [13]. But the HIP’s stability was not strongly confirmed even in investigations under postural changes. So, we have all reasons to assert that the HIP, which is located in the area of heart input under human horizontal position, will be shifted towards abdomen and, perhaps, located within lower *vena cava* under +Gz accelerations. It means that three principally different situations there may occur depending on the technical characteristics of the suit, determining the top liquid level in the “Libelle” anti-G suit.

The first situation is when the top liquid level in the “Libelle” anti-G suit is fixed on the shoulders level. Let’s analyze this hypothetical case in spite of such suit’s construction is not proposed yet. As far as the top liquid level in suit’s torso section is higher than the HIP level, we will expect that parallel to the accelerations increasing the local extravascular hydrostatic pressures will increase faster than the intravascular hydrostatic pressures. So, simultaneously to the +Gz accelerations’ increasing, these differences have to grow too. In assumption that the external pressures’ transmission coefficients into extravascular area are stable, the regional transmural pressures will decrease. Consequently, we have to observe a

decreasing of the blood volumes in abdominal and leg areas parallel to the increasing of +Gz accelerations.

The second situation is, when the top liquid level in the “Libelle” anti- G suit is fixed on the heart level. The developers of the issued “Libelle” suits proposed approximately this construction. It is not difficult to become sure in that because of HIP’s shifting down, the difference between these cases and previous one will be quantitative only: all the changes are extensively for the first case.

The third situation is when the top liquid level in the “Libelle” anti- G suit is fixed lower of the heart input level. In this case, the abdominal and leg blood volumes will first little grow simultaneously to the increasing of +Gz accelerations. Then, parallel to the increasing of the HIP’s shift toward body’s low area, the dynamics of the abdominal and leg blood volumes increasing will get slow. Theoretically, an inverse process is possible, when these blood volumes become less than in start conditions.

As far as these cases are potentially possible, we have had simulated the all three variants. As it was expected, the top protective effect was shown for the first case. Such a “Libelle” anti-G suit is able to provide the tolerance of the muscle relaxed person up to 11g for seat-back angle 12 grad. Peripheral muscle stressing increases this top value more than 2 units. We think, these simulation results might be a start point for developers to think about a creation of such suit modification.

The appropriate graphics for the mean aortal pressure (MAP), mean carotid arterial pressure (MCAP), eye level systolic pressure (PES) and central venous pressure (CVP) are presented on Fig. 21. Looking on the next figure, one can see the blood volumes’ dynamics in the same situation. As these curves are shown, the summary blood volumes are increasing both in lungs (VLungs) and in the whole thorax (VTor). But the blood volumes in abdominal vessels (VAbdom) and in legs (VLegs) are essentially decreased during acceleration’s plateau. The hand blood volume (Vhand) does not essentially change, whereas the summary head blood volume (Vhead) is increasing.

Using the middle variant analyzed above, the acceleration tolerance till to 7,5 g was reached only for the muscle relaxed person in the basic model. Changing the emotional stress from its moderate value to the high one, this top tolerance level was increased on 1g. Peripheral muscle stress enlarges this value till to 9.5 g.

The third variant mentioned above is able to provide up to 6 – 6.5 g acceleration tolerance, depending on the distance between HIP and top liquid level in suit.

Of course, these limits were higher for humans with high emotional stress, and lower – for the subjects with less emotional stress. All these results are fixed for the basic model.

These simulation examples and all our theoretic analyses are aimed to widen the empirically based imaginations about mechanism of protective effects of water filled antigravitational suits. We also hope, such analysis will help one to understand, why there were observed different effectiveness of these suits. We are sure that the approach used in “Libelle” technology really has good perspectives.

There is another aspect of this problem that is reasonable to discuss here too. We mean the possible changes in the pleural pressure under use of anti – G suits. Usually, experts assert that under condition of natural breathing the inflation of the abdominal section of suit will not increase the pleural pressure. But our model results compel us to dept in this opinion. Really, we know that under +Gz accelerations without use of suit, the diaphragm moves toward abdominal cavity and thus, parallel to accelerations’ increasing, the pleural pressure decreases. It is one of the main causes, why the lung’s blood volume increases under accelerations. Using an anti-G suit, that is able to increase the abdominal extravascular pressure, one will observe a decreasing of the diaphragm shifts. So, the pleural pressure decreasing will be less than without use of anti-G suit. The question is how much expressed may be these changes, and whether are they essential? This is one of the cases, when the computer simulations may help us by additional useful information.

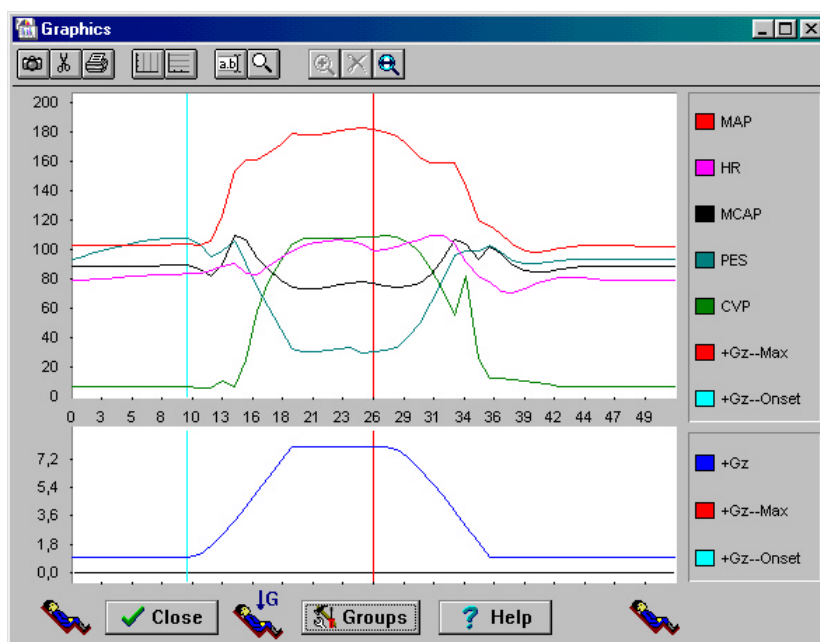


Fig.21. The dynamics of mean arterial pressure (MAP), mean pressure in carotid sinus (MCAP), systolic arterial pressure on the eye level (PES), central venous pressure (CVP) and heart rate (HR) under trapezoid profile accelerations with the acceleration and deceleration gradients 1g/s (the bottom part of the figure).

The top liquid level in the “Libelle” anti G suit was assumed located on the human shoulder level.

The muscle stress regime is Peripheral Straining and the emotional stress regime is set on its moderate position.

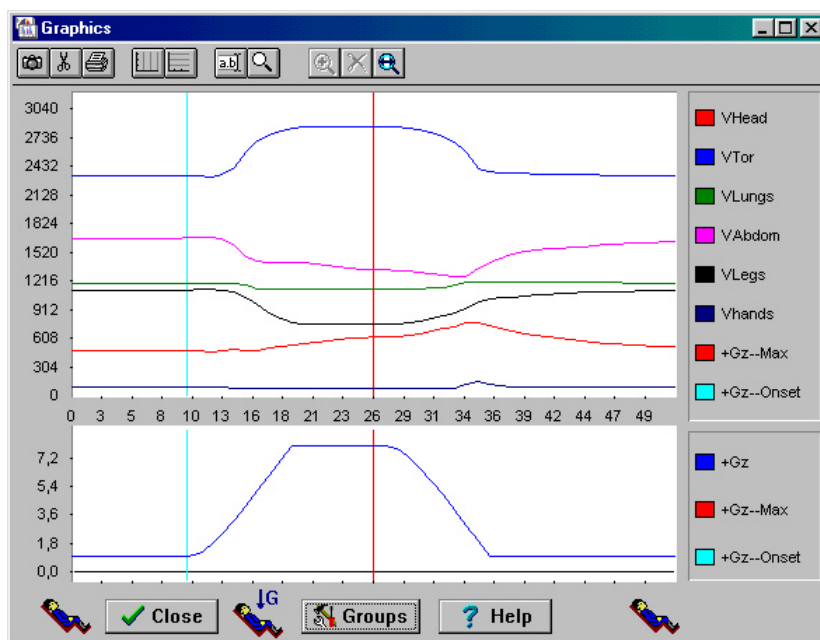


Fig.22. The dynamics of blood volumes in different body sections:

Vtor – thorax; Vabdom – abdominal; VLungs – lungs; Vlegs- legs; Vhead – head; Vhand –Hands under trapezoid profile accelerations with the acceleration and deceleration gradients 1g/s (the bottom part of the figure).

The top liquid level in the “Libelle” anti G suit was assumed located on the human shoulder level.

The muscle stress regime is Peripheral Straining and the emotional stress regime is set on its moderate position.

To be able to reflect these factors, the basic models were appropriately modified and a series of simulation experiments were done. They were aimed to investigate this problem under use of both pneumatic and ‘Libelle’ anti-G suits. The main result of these simulation experiments is that was found out a quantitative, almost linear character of one way dependence between pressure levels in abdominal and in pleural cavities. It allows us to assume that in real conditions parallel to the increasing of pleural pressure, the breathing mechanics will be compensatory transformed so that the role of chest shifting will grow up. Consequently, the vertical gravitational redistribution of blood volumes in the lung’s shares will change. Comparatively with the subject, who does not use antigravitational trousers, subjects using anti-G suit will have more lung venous pressure in his/her middle lung sections, which are the main determinants of P_{lv} , and thus the left heart input flow.

There is another aspect of anti-G suit protective effects analyze. The effects of blood volumes’ redistribution under increasing of pressures, applied to body lower part, have to be different for the muscle relaxed or stressed cases. If the legs’ and abdomen’s muscles are relaxed, the additional blood volumes that shift from the upper body areas under +Gz acceleration, will be accumulated both in skin and in muscle vessels. The use of the anti-G suit will mostly compel blood to leave skin vessels and partially to remove into relaxed deep muscles. Peripheral muscle stressing pushes out the muscle part of blood volume and thus increases the protective effect of anti-G suit. These objectives are taken into consideration to improve the vascular presentation in our model by differentiation of skin and muscle vessels. So, the actual model is able to more deep reflect protective effects of anti-G suits and muscle tension.

Usually, the pressure gradients that are used to pressure pneumatic suits abdominal and thigh sections are higher than gravitational increasing of inside hydrostatic pressures in vessels of these areas. It means the blood volumes will decrease. So, the protective effect might be less if these blood volumes still be as they were for $g = 1$. At the same time, the maximal effect of the liquid filled suit cannot be higher than recovering of volume status for $g = 1$, if HIP’s shifting is not essential.

4.2 Factors determining variability of human tolerance to +Gz acceleration.

Experts know that the top value of human tolerance to standardized profile of +Gz acceleration loading may have essential variability [2-7,9,10]. The variability is characteristic both during results comparison observed on different subjects and even in frame of different observations for one subject. Factors determining this variability were analyzed to include them in our models. Such approach will help one both to understand why the published results of different investigators have a wide disperse, and how to effectively use our models. This analysis is also aimed to determine an acceptable approach to the problem, how to optimize protections use.

Generally speaking, all anthropologic, psychological, physiological and environmental factors that theoretically may influence on the top limit of human tolerance to +Gz acceleration may be divided into two different groups, containing observable and non-observable factors. Let’s the factors that potentially might be controlled by investigator, consider here as the first factors group. Into the factors’ second group we include the factors that may be considered as causal within every observation. So, this vision platform lets us to imagine that the variability between every two observations is sooner a regular event than an exclusive one.

An additional useful condition for our analysis is the assumption that among the factors of the first group we also can mark two subgroups according to the factor’s relative role in providing of human tolerance to +Gz acceleration. The list of the major factors determining top value of the tolerance, consists of several anthropometric, psychological, physiological and environmental characteristics. For our further analysis it will be worth to consider them as input parameters.

As to the human anthropometric characteristics, the main factor is sex. Experts also have unanimity concerning diverse correlation between body length and top limit of tolerance to +Gz accelerations. But we could not find any published results concerning influences of the weight on the +Gz tolerance. At the same time, physiologists know that the summary blood volume in CVS is one of the main factors determining the level of the mean arterial pressure. From other hand, it is known that the summary blood volume is proportional to the person's weight. So, these two characteristics are essential factors, modifying the background level of hemodynamis and thus modulating the differences between subjects' tolerance to standard acceleration.

The psychological subgroup of the input parameters includes characteristics of human sensitivity, usually observed as reactivity to the expected potential dangers. It is natural, when human subject is worrying just before acceleration test on centrifuge, especially when he/she is informed about the planned to be loaded under extreme level high sustained acceleration. As to pilots' physiological conditions during real flights with combat maneuvers, pilots normally do them under high level of cathecholamines in blood. In frame of this work we do not need to know the exactly mechanism of every psychological stress-factor. But they all may be integrated and presented in our models as an input of emotional reactivity. This reactivity is able to modify physiological parameters that are characteristic for the central nervous control mechanisms of CVS without emotional stress. For example, it is well known that the levels of arterial blood pressure, cardiac output and heart rate essentially change under person's different emotional states. But at the same time it is known that these characteristics are the main among the hemodynamic variables, which determine the human tolerance to accelerations.

Certainly, to be able more exactly to calculate parameters of individual models, we need special quantitative measures both concerning the emotional stress level and additional data about deep characteristics of the CVS's control system. Usually, such information is absent among the medico-physiological measurements under accelerations. So, we could not describe this peculiarity as special equations. Therefore, we were compelled to find a compromise way to include this group of input influences on the model constants for the every subject. In the realized variant of the "Pilacel" model these differences of the subject's reactivity are presented by means of four his/her fixed background emotional states:

- Emotional stress is absent ($HR < 70$ per min.);
- Low emotional stress ($71 < HR < 80$ per min.);
- Middle stress level for every subject ($81 < HR < 100$ per min.);
- High emotional stress ($HR > 100$ per min.).

The physiological factors of individual variability of the human tolerance to +Gz acceleration include both the parameters already mentioned above and several additional characteristics. In comparison with the emotional reactivity, that even may have changes within one group of observations for one person, the physiological individuality is more stable. It may have essential changes for rather long time observations only. These changes usually are caused by the mechanism of structural adaptation. For example, the pressure – volume characteristics of the regional vessel compartment may be changed by means of increasing or decreasing of the muscle share of the vessel and/or the nervous density of regional arterioles. This kind of information is absent in usual medico – physiological investigations, but it is especially important during model creation. So, creating our models, we tried to give users an opportunity to vary several physiological characteristics to understand their role in the human tolerance to +Gz acceleration. The most of these characteristics are included into one special "Hypotheses" block. The investigators can see how important role may play the changes of these parameters.

The third group of factors, called above as environmental factors, includes such factors that usually were not registered and controlled in observations under acceleration

loading. For example, authors usually differ two muscle states only: relaxed and stressed. We have to mark that this division has never been strongly controlled. So, no one is able to present special data about intramuscular pressures that were forced in different muscle areas by every person. At the same time, our previous models have shown that even slightly changes of the muscle stress level one will be able to reach higher or less tolerance to rapid accelerations. Therefore, we have included in model of muscle stress form protection two additional interim states of muscle stress. Besides, during modeling of an antigravitational stress maneuver (AGSM), the user is able to investigate the effects of this maneuver for different top muscle stress values. He/she can also vary both the stress duration and imitate short time rapid inspirations.

We hope, the opportunities that present our model to investigators to analyze these factors, could help he/she deeper to understand why the empiric test results sometimes may essentially differ even for one person.

4.3 The problem of protections optimization

One of the problems, that was included in plan to be investigated within this phase of research, was related to the personal optimization of protection methods and tools. The theoretical analysis described in previous paragraph has already shown one side of the complicity for such optimization. But there is another side of the problem too that increases this complicity. Let's analyze these problems in detail.

We saw how much factors may determine the human tolerance to +Gz accelerations. To solve the problem of multiparametric optimization, one should know not only the potential abilities of every parameter. He/she also must know what limits he/she has for every parameter changes. Experts know, that the armchair's seat back angle reclining provides the maximal protective effect under +Gz acceleration. The next powerful protection is the AGSM. As to the developers of the "Libelle" anti-G suit, they are ensured it is able to provide protection approximately equally to the AGSM. Our simulation results are confirming this assertion. The top limit of the "Libelle" anti-G suit protection depends on additional factors such as emotional stress level, muscle stress level. The main difference between protective efficiency of the pneumatic anti-G suits and the "Libelle" anti-G suit is caused by inertia of pneumatic suits' inflation.

4.4 The problem of arterial baroreceptor reflexes under Gz accelerations.

Our early modeling results [14-17] were caused mainly by arterial baroreceptor reflexes presentation in the model of central nervous control of CVS's function. The crucial theoretic position was the assertion that the general pressor reflex there will be involved to provide the reactions observed usually. This assumption helps us to avoid increasing of the parasympathetic influences on CVS under high level aortal blood pressure. From one hand, to be able to provide high acceleration tolerance, the blood pressure will obligatory increase. Such increasing creates antagonism between reflexes from aortal and carotid receptor zones. So, the summary reflector effect becomes lower parallel to increasing of +Gz acceleration level. The reasonable question is what kind of additional reflexes or modifications of the issued reflexes interaction [19] will be assumed and described in models to be able to provide increasing of the HR and peripheral vessel tonuses.

The first reaction was to include also the reflexes from the brain basal arterial cycle. The problem is that in spite of morphologists are described mechanoreceptors in these vessels, physiologists have not investigated the role of these receptors in summary arterial baroreceptor reflex. So, this receptor zone was also included in our model of central nervous

control of human hemodynamics. The next problem was how to describe the interaction between the aorto-carotid reflexes and reflexes from brain arteriole. If we assume that these effects have to be presented in model as simple additive, we could shift the problem of depressor effects only a little towards higher accelerations. But it is not a principal solution of problem. We again will observe decreasing of HR parallel to increasing of MAP.

The new approach to this problem was a hypothesis that under antagonism of these reflexes the priority will be for the reflexes, which work to provide brain perfusion. So, the carotid and brain receptor zones will continue their pressor effects while the gain of aortal depressor reflex will become lower. Only such combination is able to explain the characteristic behavior of HR. Therefore, our last model modification of BCM and ARM presents such reflex interaction.

5 RESUME

The actual “PILACCEL” software, that is the next step on a way of the adequate theoretical technology creation, able to provide physiologist-researchers during their investigations in area of human physiology under piloting acceleration, is developed. The theoretical basis of the actual “PILACCEL”, calculation algorithms and software are essential improved during this development phase. Now the “PILACCEL” is able to substitute the real object - prototype for the several hypothetic situations, in which investigator is interested in, but there are serious economical and risk reasons to model them by usual empiric way. At the same time, there is a necessity to search a way, how to reach a satisfactory validation of models. This doubt mostly relates to the individual models, because of the used individualization technology is the first approach only. Certainly, the main principles used during individual model creation are physiologically well based. But nobody in World is possessing now by data, enough one to be sure that all these principles still be right for the high sustained acceleration conditions. To create a more adequate individualization technology, experts need special experiments, planed together by modelers and physiologists.

The “PILACCEL” may be used by student – pilots to better imagine the physiologic background of accelerations, the real protective opportunities of different protections and to optimize the training process.

As creators, we see a good reason to continue our collaboration on the direction of space flights. Using this development experience and knowledge in area of cardiovascular space physiology, we could create special computer based technology that will cover all acceleration and weightlessness problems, arising under modern airspace flight. Besides, we could take experts' attention on the way, how to improve the protective efficiency of water filled antigravitational suits: the top level of torso water tubes has to be longer and reach shoulders.

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PART 2:

THE USER MANUAL OF THE “PILACCEL” SOFTWARE

The user manual describes the functionality of the “PILACCEL” software and consists of general information about the user interface. It also presents information necessary one to be able to set actual model parameters, to make a computer simulation experiment, and also to analyze the results of experiment. The detailed information about theoretic basis of the “PILACCEL” one can find in the first part of this report. The variant of the user manual, that also contains a short basic information about mathematical models, is incorporated into “Help” block of the “PILACCEL” software.

1 THE FUNCTIONALITY OF THE “PILACCEL” SOFTWARE

The “PILACCEL” software is special information technology, created to simulate and theoretically investigate the main dynamic processes in human cardiovascular system under piloting Gz accelerations. It is based on a complex of quantitative mathematical models that consist of about 1200 differential and algebraic equations.

The “PILACCEL” is mainly oriented to the physiologist-researcher, who is interested in a deep knowledge concerning mechanisms and limits of human hemodynamics under high sustained accelerations. He/she may use this knowledge to develop more effective protective methods and tools or optimize the use of existing tools. In addition, the “PILACCEL” may be used as an educational tool able to show students – pilots the main hemodynamic effects, that are causing by piloting accelerations with – and without use of different protections. In this case, the “PILACCEL” may help students to optimize training process aimed to provide human tolerance to the high sustained accelerations.

The “PILACCEL” is autonomic software. It provides experimenter by computer simulation experiments that will be prepared and executed in an interactive regime by means of user interface. There are two different approaches to the simulation experiments.

Within the first approach, the “PILACCEL” gives one an opportunity to simulate hemodynamics of the “mean” man. This model is called the basic one. The user able to set parameters of acceleration loading regime and protections, while all inside parameters of hemodynamic models are fixed. The only change of the basic model characteristics is possible through the use of the special “HYOTHESES” block. It allows user to set four different values of the general stress level that characterize the human background emotional state just before acceleration loading. In addition, the user is able to vary several characteristics of the central nervous reflector control mechanisms that control the heart pump function and the tonus of the arterial and venous vessels.

Within the second approach, the “PILACCEL” gives one also some additional opportunities to simulate the individual hemodynamic effects of the accelerations for the persons, using their characteristics such as sex, age, body weight and length, health. After the user sets these data for the person, the algorithm of the “PILACCEL” provides automatically changes of the basic model parameters, and creates the individual model.

The “PILACCEL” presents the results of the every simulation experiment in three different forms: a table form; a graph form; a form that is named as “Experiment protocol”. During use of the first two forms, the “PILACCEL” creates its own presentation forms. Besides, there is another possibility to present results by means of the incorporated Excel module activation. The “Experiment protocol” is automatically generating for the every simulation experiment, and the user may write his/her comments down and print or save it. A special program provides results comparison in graph form for the every two chosen simulation experiments.

Both the experiments’ options and their results may be saved as special files that may be loaded whenever the user would like.

2 GENERAL INFORMATION ABOUT THE USER INTERFACE: THE MAIN SCREEN FORMS

The first screen form of the “PILACCEL” software is a combination of pictures, aimed to create an imagination about the professional and problem area, where it may be used. This form appears every time when the user is first running the software. It will disappear every time, when one clicks the mouse or touches the keyboard. In versus case, it will disappear 30 sec later. The next screen form that is the main screen form of the “PILACCEL” software becomes visible instead of the disappeared first form.



Fig.25. The first screen form of the “PILACCEL” software’s user interface.

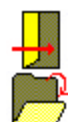
2.1 The main screen form “Piloting Accelerations”

The main screen form (see Fig.26) serves all user operations by means of two alternative ways. The standard way of the user access to the operations is through the four acceptable control options (**File, Experiment, Results, Help**). Their inside functions are clearly enough described and the user needs not any additional comments. The second way of the screen form use is a direct access to the 12 speed buttons that are providing the same functions. A short information about the function of every speed button is presented in the appropriate hint - mini help-window that appears every time when the user points the chosen button by mouse pointer.



Fig.26. The main screen form of the “PILACCEL” software’s user interface.

Speed-buttons’ hint information and functions are following:



Exit – it is used to quit the “Pilacel”



Open options – gives one an opportunity to use the Experiment options, saved early.



Save options – allows saving of the current Experiment options.



Experiment options – allows settings of new Experiment options.



Hypothesis – allows the settings in special screen form window “Hypotheses”



Calculate – allows starting of calculations with the actual model’s sets.



Graphics – gives one an opportunity to use the graphic form of the simulation experiment’s results presentation.



Result table – gives one an opportunity to use the table form of the simulation experiment’s results presentation.



Export to Excel – gives one an opportunity to use the incorporated Excel module to present the results of the simulation experiment in graphic and table forms.



Remember for comparison – will be used by the user every time if he/she wants to compare the actual results with results of the second experiment. To remember the results of the current simulation experiment, this speed button will be pushed before to change the experiment options and to execute a new simulation experiment.



Compare with the previous remembered – allows the comparison of the actual simulation experiment’s results with the experiment that was already done and remembered.

Notes results of experiment – Pushing this button, one can form and open special Experiment Protocol that allows the user add his/her comments concerning the simulation experiment.

2.2 The screen form “Experiment Parameters Setting”

The next screen form with a general functional name “Experiment Parameters Setting” is shown on Fig.27. As one can see, it contains of four red tabs:

Models

Acceleration Profiles

Protections

Interface options

They are assistant forms to provide all settings necessary to execute a computer simulation experiment. More detailed information about these interface forms will be presented below.

The screenshot shows a window titled "Experiment Parameters Setting" with four tabs: "Models", "Acceleration Profiles", "Protections", and "Interface options". The "Acceleration Profiles" tab is active. On the left, under the "Models" section, there are five radio buttons: "Linear Increase", "Linear Increase Plateau", "Trapeze Profile", "Push-Pull effect" (which is selected), and "Arbitrary Profile". On the right, there are five input fields: "Alpha=" (1.00), "Beta=" (1.00), "Gmax=" (8.50), "T=" (10), and "T1=" (3). To the right of these fields is a graph of acceleration G versus time t. The graph shows a red curve that starts at 1, dips to -1, then rises to a peak labeled G_{max}. The rising part is labeled T, and the falling part is labeled T₁. The peak is also labeled α and β. The graph is enclosed in a dashed red box.

Fig.27. The screen form “**Experiment Parameters Setting**” of the “**PILACCEL**” software’s user interface. It contains of four independent parts that are serving the experiment preparation. Now the option “**Acceleration Profiles**” is active and the “**Push-Pull effect’s**” profile is chosen.

2.2.1 Models

There are two different models: basic and individual. To choose the basic model, one should activate the appropriate check-box only. To choose the individual model, one should first activate the appropriate check-box. It transfers the connected field on the right side into black of color. Now the user should set in causal order the actual values of person’s name, age, weight and length of body. It is necessary to activate both one of two health states: “Normal” or “Weak”, and also “Male” or “Female” boxes.

The check-box “Regular flights” is common for the both types of human models. Its’ activation means that the pilot is flying regular, while the empty position means irregular flights.

Experiment Parameters Setting

Models | Acceleration Profiles | Protections | Interface options

Models

☐ Basic Model

☒ Individual Model

☒ Regular flights

Person

Comments:

Sex: ☐ Male ☐ FeMale

Age: years (18-60)

Height: cm (150-200)

Weight: kg (50-120)

Health: ☒ Normal ☐ Weak

☒ OK ☐ Cancel ☐ Help

Fig.28. The screen form “**Experiment Parameters Setting**” of the “**PILACCEL**” software’s user interface.

The option “**Models** ” is serving the model type choice. Now the “**Individual model**” is chosen.

2.2.2 Acceleration Profiles

This screen form allows user to choose one of four fixed profiles, or to build special arbitrary profile that is absent in standard “Pilacel”. All constructed profiles may be saved by user and used by him/her in future directly loading them. To build an arbitrary profile, the user should first click 2 times on the appropriate white field. It activates the new screen form named as “Drawing G-Profile”. The profile building procedure is maximally simple: it is enough to consecutively set points on the field by mouse. All assistance information about G-profile is also presented in this form. The user can more preciously set its characteristics by means of special table.

The fixed profiles are:

- **Linear**
- **Linear to Plateau**
- **Trapezoid**
- **“Push-Pull effect”**. (This one was presented on Fig.27.)

Choosing one of these regimes, the user can see it graphically. Besides, the acceleration (Alpha) and deceleration (Beta) gradients, the plateau level (Gmax) and duration (T or T1) are also available to set.

Fig.29. The screen form “Experiment Parameters Setting” of the “PILACCEL” software’s user interface.

The option “**Acceleration Profiles**” is serving the choice of the acceleration profile. Now the option “**Trapeze Profile**” is chosen.

In case the “**Arbitrary Profile**” is chosen, the special screen form (see Fig.30) will be used to build the Arbitrary Acceleration Profile.

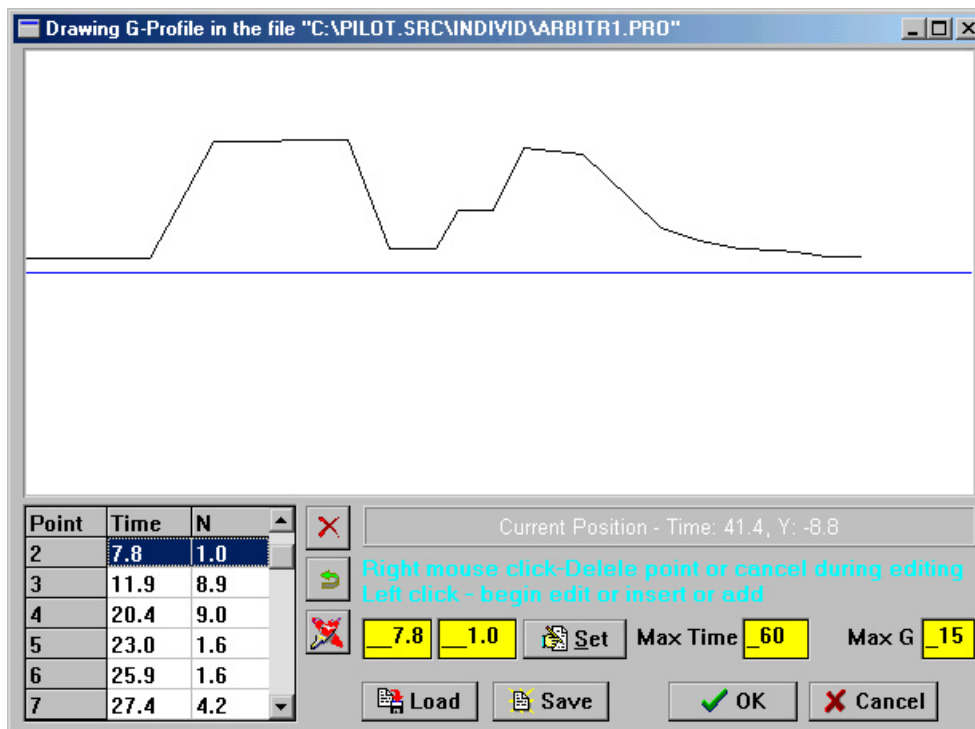


Fig.30. The screen form “Experiment Parameters Setting” of the “PILACCEL” software’s user interface.

The special screen form is serving the **Arbitrary Acceleration Profile** construction.

2.2.3 Protections

Four types of protections are available to use in the ‘Pilacel’:

- **Muscle stress;**
- **Seat-angles;**
- **Breathing Regimes;**
- **Anti-G Suit.**

They all present to the user several variants of inside parameters setting.

Using block “Seat-angles”, one can set three angles, visually presented on the pilots’ figure. There are five positions of pilot illustrating different seat-back angle situations. This figure is also able to demonstrate anti-G suit types or its’ absence.

Using block “Muscle stress”, one will be able to choose one situation from the list consisting of four positions:

- **Relaxed**
- **Hand gripping**
- **Peripheral Straining**
- **AGSM .**

Excluding the first position, three others need an additional parameter setting. It is the G- Threshold, that will set the reached G-level in model to start the muscle stressing.

Choosing the AGSM, one should set three additional data that have to characterize the AGSM’s parameters as: Maximal pressure (in mm Hg), reached by AGSM, its’ duration (in sec) and duration of the short inspiration (in sec).

A special warning appears to note that in ‘Pilacel’ AGSM cannot be realized in the regime, when hemodynamics’ breathing pulsations may be indicated, or the arbitrary breathing is set.

To set protection using the box “ Breathing Regimes”, one should remember that three breathing regimes are available:

- **Natural**
- **Positive Pressure**
- **Arbitrary.**

In fact, only the last two regimes can be considered as protective. To set the regime “Positive Pressure”, one should first choose it: the special mini-window will appear. It contains a figure, demonstrating that breathing pressure will linearly increase with the gradient “Alpha” starting from several Gt –threshold, and will stop its increasing after the Pmax is reached.

To set the arbitrary breathing profile, one should first double click on an appropriate white field. This procedure opens a special window similar to the window used to build arbitrary acceleration profile. After the necessary breathing profile is built, it may be saved and loaded.

The block “Anti G Suit” gives one an opportunity to choose one situation from the three possible situations:

- **Absent**
- **Pneumatic**
- **Libelle.**

Choosing the first position, one can simulate the situation when the pilot is not worn in anti-G suit.

Choosing the second situation, one can simulate the standard pneumatic anti-G suit using. This suit is presented in the model consisting of three (abdominal, thigh and shank) sections. As far as different suits within this class of suit may have different characteristics,

the user must also set several additional data. He/she will confirm that the section is presented, and then input the percent of covering for every section. The gradient of pressure inflation may be set using one of two units (mm Hg or Psi). The second value will be indicated automatically. The user has to set also the G-threshold, pressure inflation will start from which.

Fig.31. The screen form “Experiment Parameters Setting” of the “PILACCEL” software’s user interface.

The screen form is serving the setting of the **Protections**.

This screen form has several modifications, serving both the choice of concrete protection method and its parameters’ setting (see Figures 31A, 31B, 31C).

Sections	% of covering
<input checked="" type="checkbox"/> Abdomen	40
<input checked="" type="checkbox"/> Thigh	75
<input checked="" type="checkbox"/> Shank	77

Fig.31A. The screen form “Experiment Parameters Setting” of the “PILACCEL” software’s user interface.

The screen form is serving the setting the choice of the **anti-G suits**.

Experiment Parameters Setting

Models | Acceleration Profiles | Protections | Interface options

Muscle stress: Relaxed (dropdown menu open showing: Relaxed, Hand gripping, Peripheral Strain, AGSM)

Breathing Regimes : Natural

Seatangles: A=30, B=-10, C=55

Anti G Suit : Pneumatic

G-Threshold: 2.00

Sections % of covering:

Sections	% of covering
<input checked="" type="checkbox"/> Abdomen	40
<input checked="" type="checkbox"/> Thigh	75
<input checked="" type="checkbox"/> Shank	77

Pressure Gradient: 55 mm Hg, 1.00 Psi

OK Cancel Help

Fig.31B. The screen form “Experiment Parameters Setting” of the “PILACCEL” software’s user interface.

The screen form is serving the setting of the **Muscle stress regime** choice.

Experiment Parameters Setting

Models | Acceleration Profiles | **Protections** | Interface options

Muscle stress: Peripheral Str

G-Threshold: 2.00

Breathing Regimes : Positive Pressure

Graph: P (mm Hg) vs. G. Points: Pmax, Pt, Gt, Alpha.

Parameters: Pmax=60 mm Hg, Pt=15, Alpha=15, Gt=4.00

Seatangles: A=60, B=-10, C=55

Anti G Suit : Pneumatic

G-Threshold: 2.00

Sections % of covering:

Sections	% of covering
<input checked="" type="checkbox"/> Abdomen	40
<input checked="" type="checkbox"/> Thigh	75
<input checked="" type="checkbox"/> Shank	77

Pressure Gradient: 55 mm Hg, 1.00 Psi

OK Cancel Help

Fig.31C. The screen form “Experiment Parameters Setting” of the “PILACCEL” software’s user interface.

The screen form is serving the settings of the **Armchair’s angles & anti-G pneumatic suit & Positive Pressure Breathing Regime** choice while the **Muscle Stress regime** is set **Peripheral Straining**.

2.2.4 Interface options

The function of the screen window “Interface options” is to correct several software parameters according to the user actual interests.

This window includes options that are presented into two groups. The left side located group, named as “Interface Options”. In the right side of window is located “Run Time Controls”.

In fact, all presented functions are clearly enough described and one can understand their meaning without any additional comments.

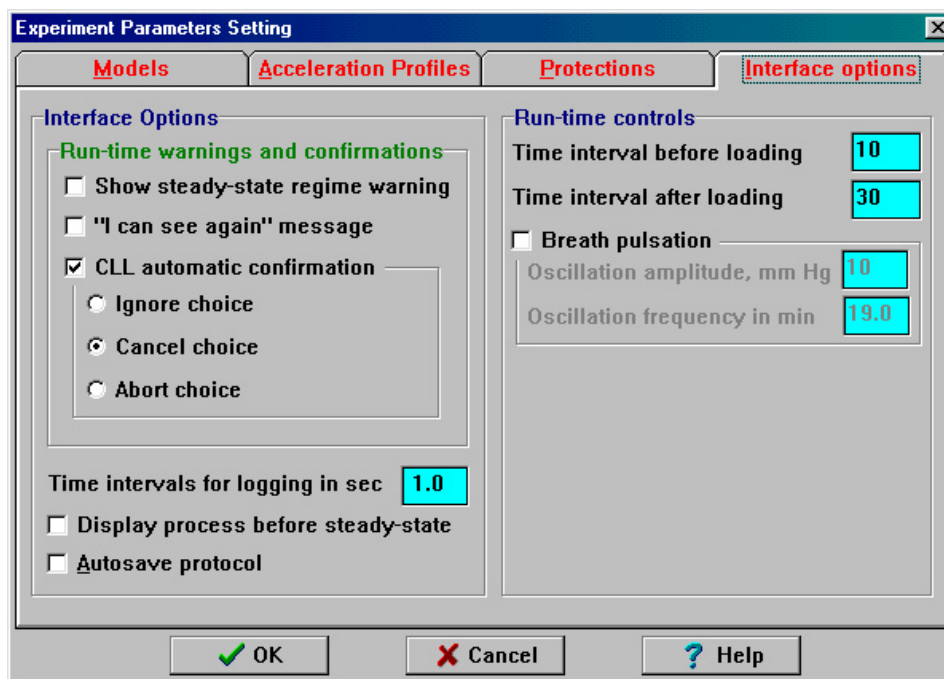


Fig.32. The screen form “**Interface Options**” of the “PILACCEL” software’s user interface.

The screen form is serving the setting of the assistance settings.

2.3 Hypotheses Selection

This window presents to user an opportunity to do the investigations with the help of the “Pilaccel” a little more physiologically interesting. But, to be able to use these opportunities, the user will know the theoretical concept of models, presented above.

This window consists of two independent blocks: “Nervous Control”; “Emotional Stress”.

The block “Nervous Control” includes presentation of aortal and carotid baroreflexes, controlling the HPF and the vessel tonus, and also the Bainbridge reflex, controlling the HPF only. The nervous control of the cardiovascular system is in function when the check-box is activated only. According to the modeling concept, the reflexes are presented by their gains (Coefficients) and time constants. Within this block, one can switch of the nervous control for every presented hemodynamic parameters. Besides, one could simulate relatively increasing or decreasing rates of aortal or carotid reflexes. The user is also able to change the nominal values of regulators’ characteristics. Information about acceptable limits will appear every time the mouse occurred on the appropriate field.

The block “Emotional Stress” allows one to investigate the role of the emotional stress in the mechanism of acceleration tolerance providing. There may be four fixed situations for the emotional stress simulation:

1. Absent – the check-box is not activated;
2. Low stress. The check-box is activated and the left position is chosen;
3. Moderate stress. The check-box is activated and the middle position is chosen;
4. High stress. The check-box is activated and the right position is chosen;

An assistance information, aimed to help one to understand this graduation, is presented in the appearing mini-windows, that presenting the value of the heart rate for the background situation (just before acceleration loading).

Hypotheses Selection

☒ **Nervous Control**

Baroreflexes ratio

☒ Aortal 0.40

☒ Carotid 0.60

Vessels

NervContrRes Coef. 1.00

NervContrD Coef. 1.00

NervContrU Coef. 1.00

Time Constant [s] 15.00

Heart

NervContrHeartRate Coef. 1.00

NervContrInotropism Coef. 1.00

Bainbridge's Reflex

Heart Rate Coef. 1.00

Inotropism Coef. 1.00

Time Constant [s] 2.00

☒ **Emotional Stress**

Background Stress Level

☐ Low ☒ Moderate ☐ High

OK Cancel

Fig.33. The screen form “ **Hypotheses Selection** ” of the “PILACCEL” software’s user interface.

The screen form is serving the additional settings of the PM’s parameters.

3 HOW TO EXECUTE A SIMULATION EXPERIMENT

To prepare a simple simulation experiment, one should use following screen forms:

- **Models.**
- **Acceleration Profiles**
- **Drawing G-Profile (Arbitrary Acceleration Profile Creation only)**
- **Protections**
- **Interface Options**
- **Hypotheses Selection**

In each of these screen forms the user able to set the actual value of model parameter. The software provides saving of actual values for the next experiment too. Therefore, before to do the next experiment, it is enough one to change the value of the parameters, in which role's investigation he/she is interested in.

Parameters' setting is available when the appropriate field is black color. There are two type of model parameters: simple and complex. The simple parameter is presented in the user interface by means of simple check-box. To set the simple parameter it is enough to transfer the appropriate check box from gray color to black color. Every group of complex parameters is functionally connected in one model. So, they are visually included into one group on the interface. To set model parameters, relating to chosen functional group, one should first activate the checkbox, as it was done for the simple case. The values of the model's complex parameters may be set after the field already is black color.

After all preparations, using the forms of **'Experiment Parameters Setting'**, have been done, the speed button "GO", located on the central area, will be activated. It starts the calculation process that will be confirmed by the special window named **'Calculations in Progress...'**.

The user should note that because of large system of equations, the calculation process requires several times, depending on a computer power. All run time service information will appear according to sets in **'Interface Options'**. When the calculations are completed, the user will again see the screen form **'Piloting Accelerations'**. Take attention, the 5 speed buttons from 6, located the right side from the "GO", now already are of black color. It means, the user has an access to these buttons to analyze the results of the simple simulation experiment.

Special Notes:

For the first several seconds of calculations the time indicators do not work. It's OK! These calculations are necessary models to reach a physiologically reasonable starting regime of hemodynamics!

The "Pilacel" software automatically saves the last settings!

4 HOW TO ANALYZE THE RESULTS OF THE SIMULATION EXPERIMENT

The main way to analyze simulation experiment results is to use graphics. Besides, one can use a table presentation of data. Both these opportunities may be realized in two ways: first, using simple forms; and second, using incorporated Excel module. If one wants to use the last format, he/she should open the appropriate window and follow to the incorporated instructions.

4.1 The run-time interaction with the “Pilaccel”.

The main calculation algorithm is tuned to break accelerations and to transfer models to the regime of decelerations after the CLL-event occurs. The software fixes this event at once the systolic pressure has felt below 30 mm Hg, and the time, equal 1,5 seconds, and necessary to recognize the CLL, is passed. The user is able to receive an information about CLL, if the appropriate position on the screen form “Interface Options” is activated. It is also possible to ignore the CLL-event and to continue the regime of acceleration loading. In this case, one can see what critical changes will occur in hemodynamics. However, the software informs the user when the full block of brain perfusion is reached, and transfers the calculation regime to the decelerations until $G_z=1$.

There is an additional opportunity to save the results of every experiment as special “Experiment Protocol”. It indicates all important experiment data. It may be added by experimenter comments, and may be saved both in file and in a hard copy form.

There are four screen forms able to show different variables’ graphics. They all have a nickname “**Graphics**”. One to be able to select the variables group, he/she first must activate the bottom located position “**Groups**”. It gives an access to the screen forms named “**Select Variables Groups**”. There are four such groups. Each of them explains the meaning of variables and their units. There are standard active or passive positions. The user is able to change these regimes. The only action, he/she should do, is to change the actual regime of the chosen checkbox.

By activation one of the **Groups**, the user can watch graphics, illustrating both G_z and variables’ dynamics.

There are some additional software opportunities too. To see the variable value, one can set the mouse position on the appropriate line and double click. The small window will appear with the necessary information. If one wants to amplify graphics, he/she should use speed buttons “Zoom in” and “Zoom out”. It is also possible to set horizontal and/or vertical lines using appropriate speed buttons.

There also is an opportunity to graphically compare the results of two simulation experiments. To do such comparison, one should first remember results of this experiment using the fourth right speed button. One can see that the color of this button transfers to the black color too. Then, changing some of parameters, the user should do the next simulation experiment. Just after the new experiment is completed, one can compare these experiments by clicking the mentioned speed button. The “Pilaccel” provides the comparing process for all the list of variables. One should choose the variable, he/she is interested in, and click OK. During the comparison of two experiments, the differences between this and previous experiments are presenting on the right field.

It is possible to obtain different combinations of the proposed mechanisms by activation or deactivation of the appropriate central nervous control mechanisms or their several CVS-parameters. One can also set different values for the coefficients and for the time constants of heart’s and/or vessels’ control parameters. In this case the following parameters of nervous activity will be available:

- heart rate (F) and heart inotropism (k)- to control of heart pump function ;
- the arterial resistance (R), the venous volumal rigidity (D) and vessels’ unstressed volume (U)- to control of vascular tonus parameters.

If the value of a coefficient is 1, it means that we have a regime of physiological norm (excluding the coefficients for aortal and carotid baroreflexes that normally equal 0.5). Mini help-windows, located below, inform user about limits of coefficient changes.

This window also allows one to set the level of nervous stress of a person just before Gz onset (assistance information about the heart rate value for each of the three variants of stress level is also presented in the mini help-windows located below).

Such a combination of settings is named “Experiment Option”.

After all these settings are performed, the user can press the button “GO” and start the computer simulation experiment. Information confirming the start will appear on the screen in the form of special window, “Calculation in progress”. This window contains a running circle on its left side. The right field informs about running time as well as the running and maximal Gz levels. When the user starts it for the first time, and every time afterwards that some of seat-angles have changed, he/she should first wait a fixed time (30s) until a new stable hemodynamic regime for the starting (phone) position is reached. The appropriate information “Hemodynamics has reached its steady-state regime for G onset” will appear on the screen. The calculation, also visually indicated by the running circle, will stop. Then the user should press “OK” and go on.

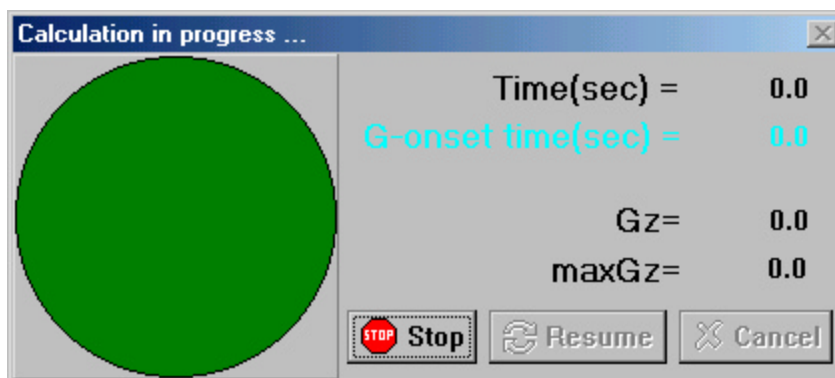


Fig.34. Screen forms after the calculation has started.

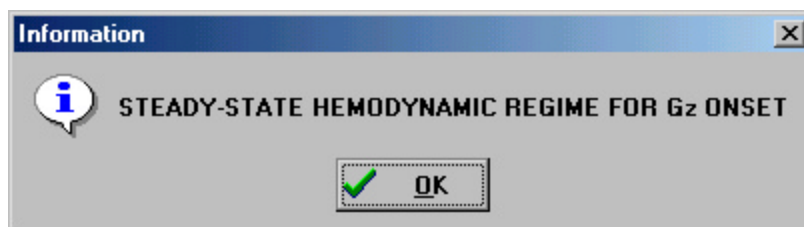


Fig.35. An assistant screen form to confirm that the program is ready to start acceleration loading.

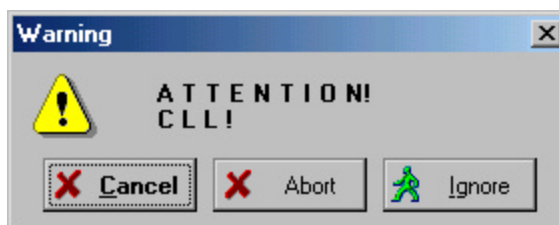


Fig.36. The screen form that shows the stopping of calculations due to CLL.

Calculations will continue until all the planned acceleration profile is completed, or the systolic arterial pressure in eyes fall below 30 mm Hg. This situation is defined as a CLL (central light loss). When CLL takes place, “PILACCEL” informs the user of it by placing a warning on the screen, presented on Fig.36. Having received this warning, the user should select one of three buttons. If “Cancel” is chosen, the acceleration’s increasing is interrupted

and the user is transferred to the regime of deceleration. In this case, the recovering processes in hemodynamics starts. So, the dropped arterial pressure begins to increase again. When it becomes higher than 30 mm Hg, “PILACCEL” will show this with the next screen form (see Fig.37).

If the user chooses the “Abort” button, they break all calculations and the “PILACCEL” returns to the starting regime.

The scenario will differ when one has pressed the button “Ignore”. It means the program will continue increasing of accelerations level in spite of the warning about CLL. In this case some time later the blockade of brain perfusion might occur because the arterial pressure continues to decrease. The appropriate information will appear as it illustrated on fig.39.

According to the biochemical mechanisms, approximately 1.5 sec after the brain perfusion is blocked the human consciousness will be lost (LOC). If this situation happens, one must break the loading process and start the deceleration.

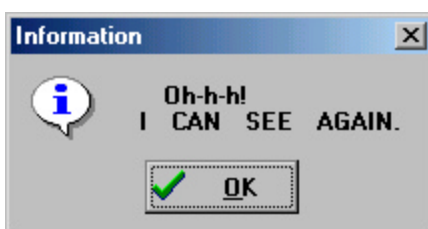


Fig.37. The screen form that appears some time later after “Cancel” is chosen.



Fig.38. Screen form indicating the end of CLL and hemodynamics' recovering .

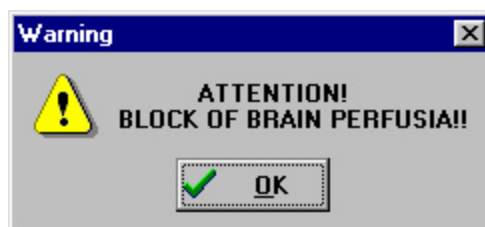


Fig.39. Screen form warning the blockade of brain perfusion.



Fig.40. Screen form warning the loss of consciousness (LOC).

For all of the three chosen regimes the “PILACCEL” fixes parameters for the top level of accelerations (see Fig.38). These parameters are also indicated on the special graphic forms in the presentation of results (one can see this as a vertical red line on appropriate figures in the next chapter.).

4.2 How to use Results of Simulation Experiment?

The software “PILACCEL” provides users with three forms for watching and analyzing the Results of Simulation Experiment: Graphic, Table and Special Experiment Protocol. A special Excel-presentation of graphics and table is also available. The appropriate speed buttons are located to the right of the button “GO”. These buttons can not be activated at the start of the experiment (they have green color). They become accessible just after calculations are done.

If the user wants to compare the results of two simulation experiments, they should first save the results of the current experiment using the button located just to the right of “Table”. Then, conducting the next experiment, one can use the button “Compare”. It is possible to compare only two graphics of one variable. The list of these variables is available in special window “Select Experiments for Comparison”(see Fig.41A).

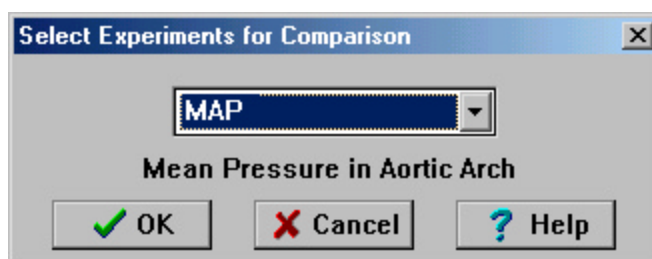


Fig.41A. The special window serves selection of variables for comparison. Now MAP is chosen.

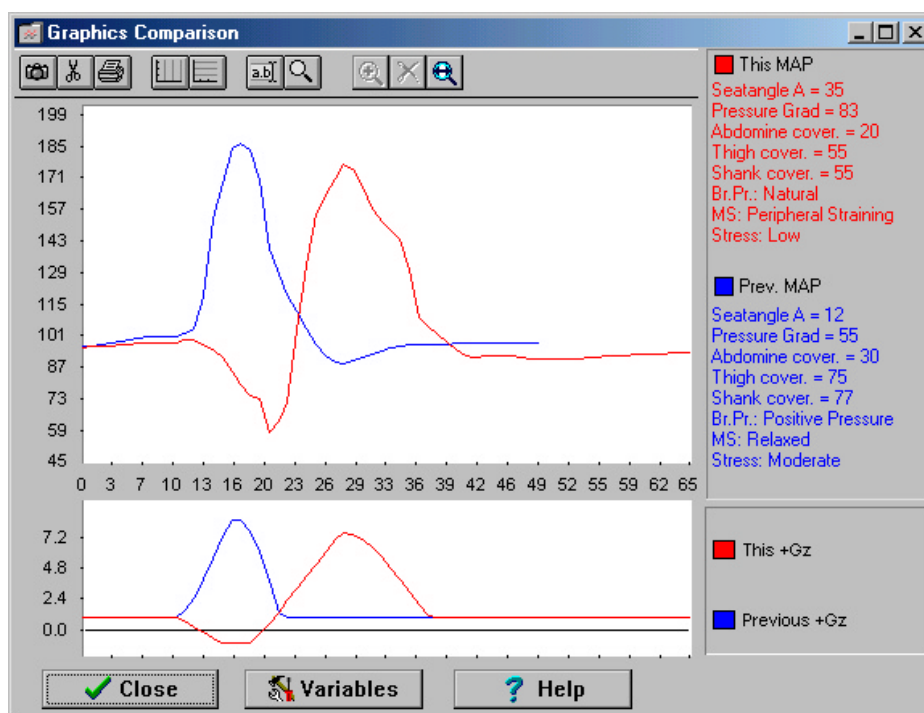


Fig.41B. A illustration of graphs' comparison for MAP.

All differences between two simulation experiments are presented on this window

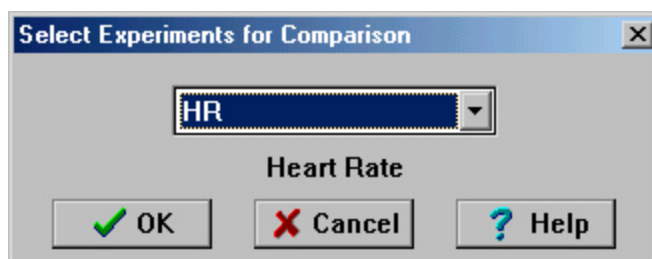


Fig.42A. The screen form is illustrating the case when HR is chosen.

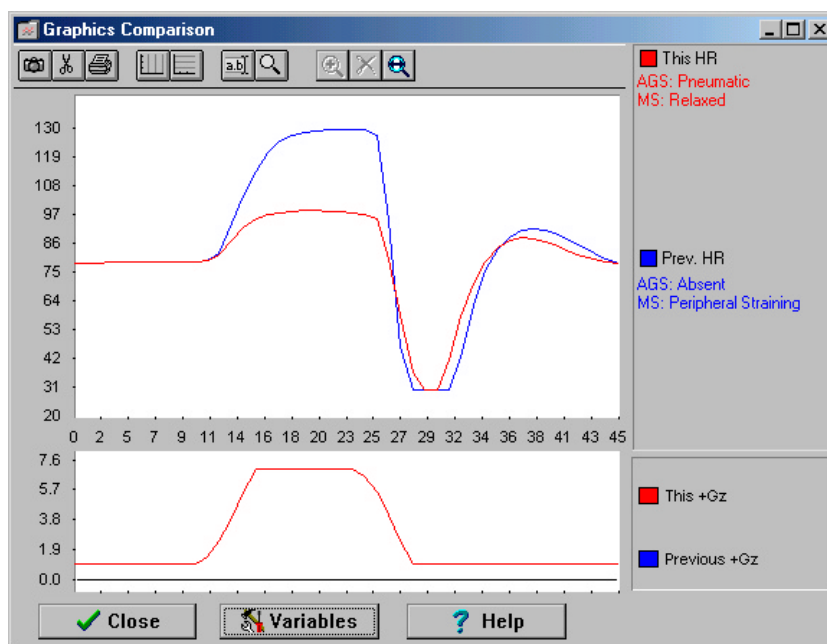


Fig.42B. Graps' comparison for the heart rate.

One can also see the application of a coordinate net.

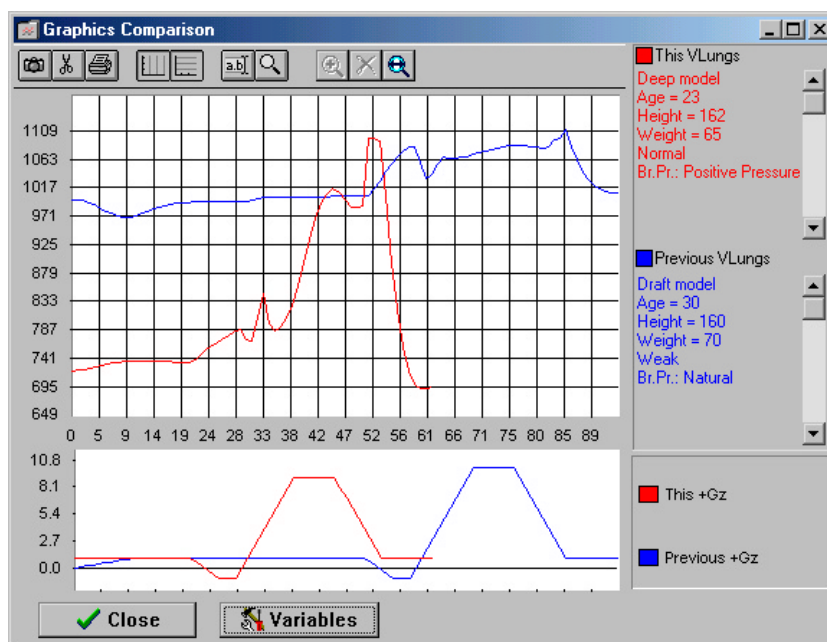


Fig.43. Graps' comparison for the lung blood volume using a coordinate net presentation.

The screen form “Select Variables Groups” is created to choose one of the four variables group for actual analysis. These groups have different inside. Fig.44 illustrates variables includes in the first group. One can see that every variable has its short name (abbreviation) and full name, including its unit. The user can change the quantity of variables by clicking on the appropriate check-box and changing its status.

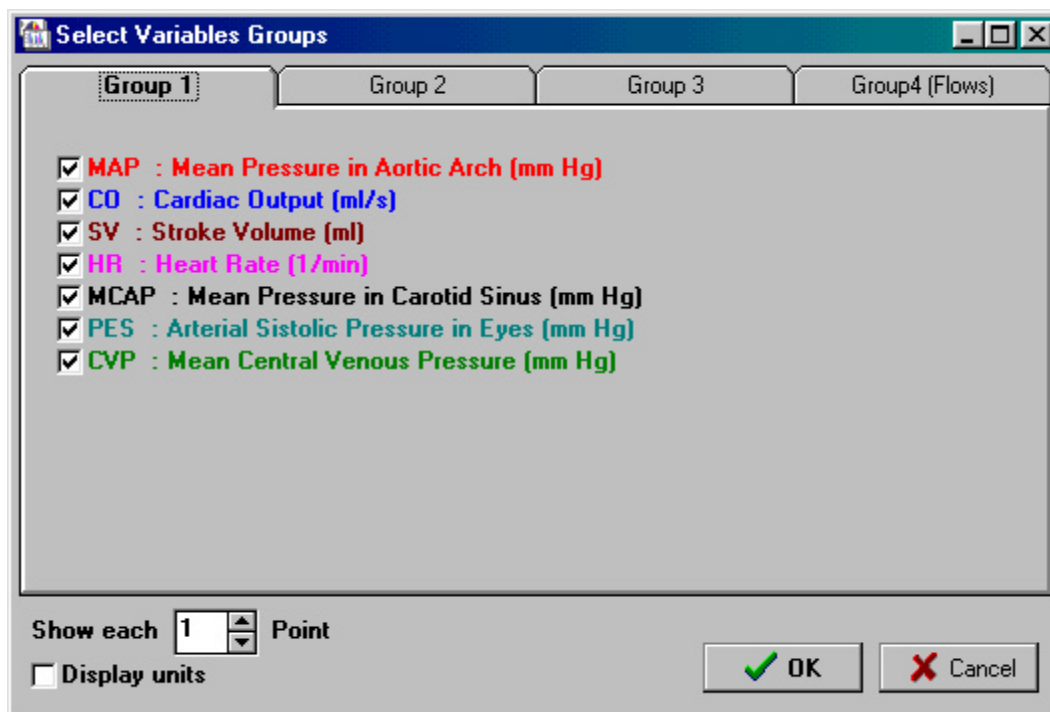


Fig.44A. The general form of screen window “Select Variables Groups” and inside for the variables of Group1.

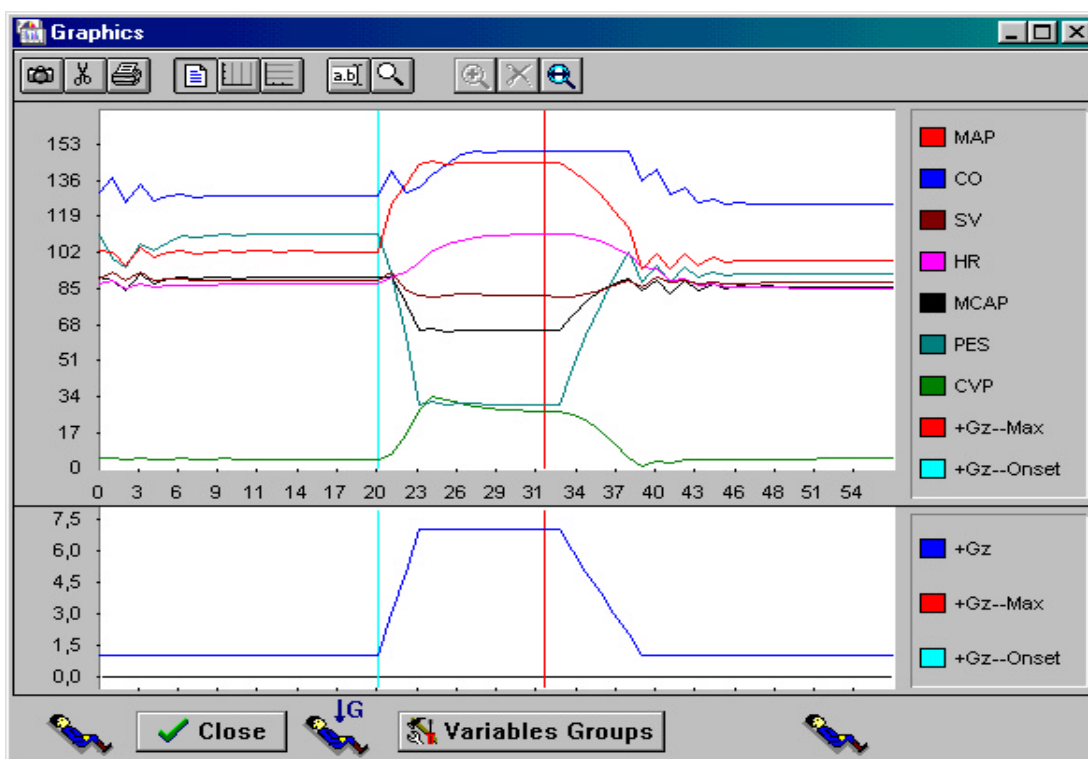


Fig.44B. An example of variables’ graphic presentation for the Group1.

The inside structure of variables for the Group3 and an example of their graphics are presented on Fig.45A and Fig.45B respectively.

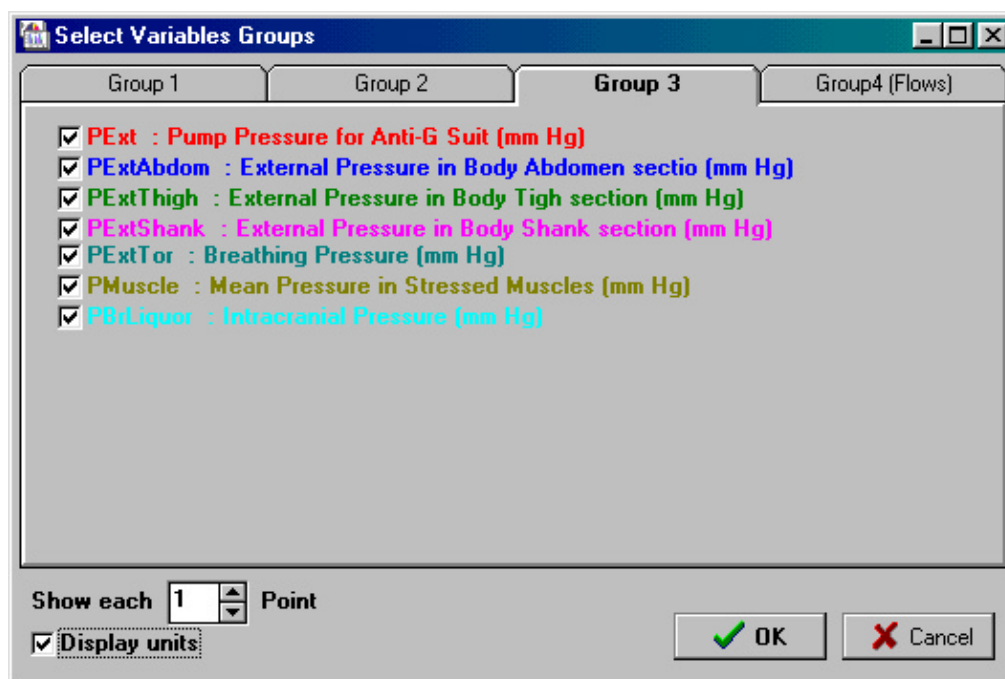


Fig.45A. The general form of screen window “Select Variables Groups” and inside for the variables of Group3.

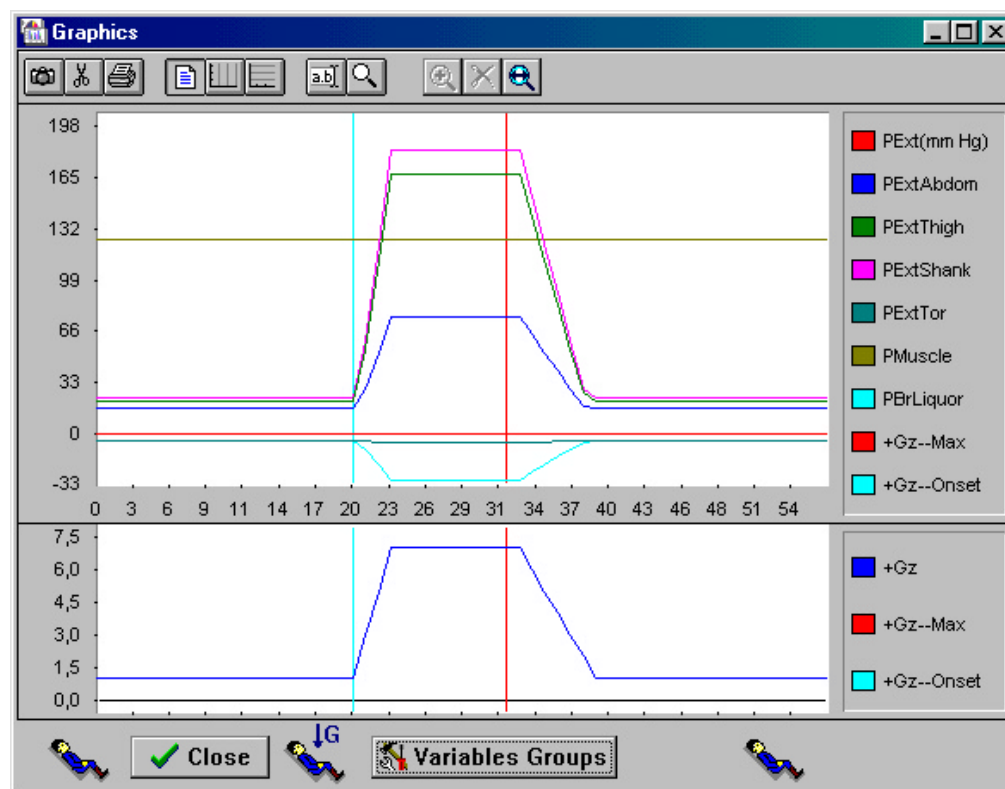


Fig.45B. The screen window illustrates an example of results' indication for the variables of Group3.

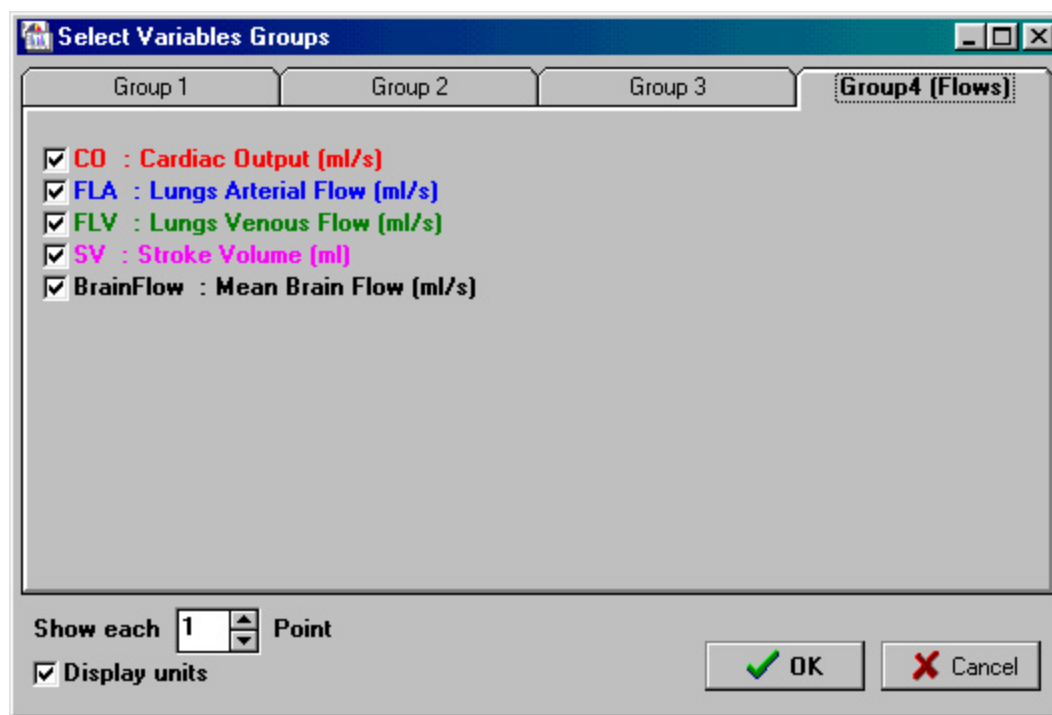


Fig.46A. The general form of screen window “Select Variables Groups”and inside for the variables of Group4.

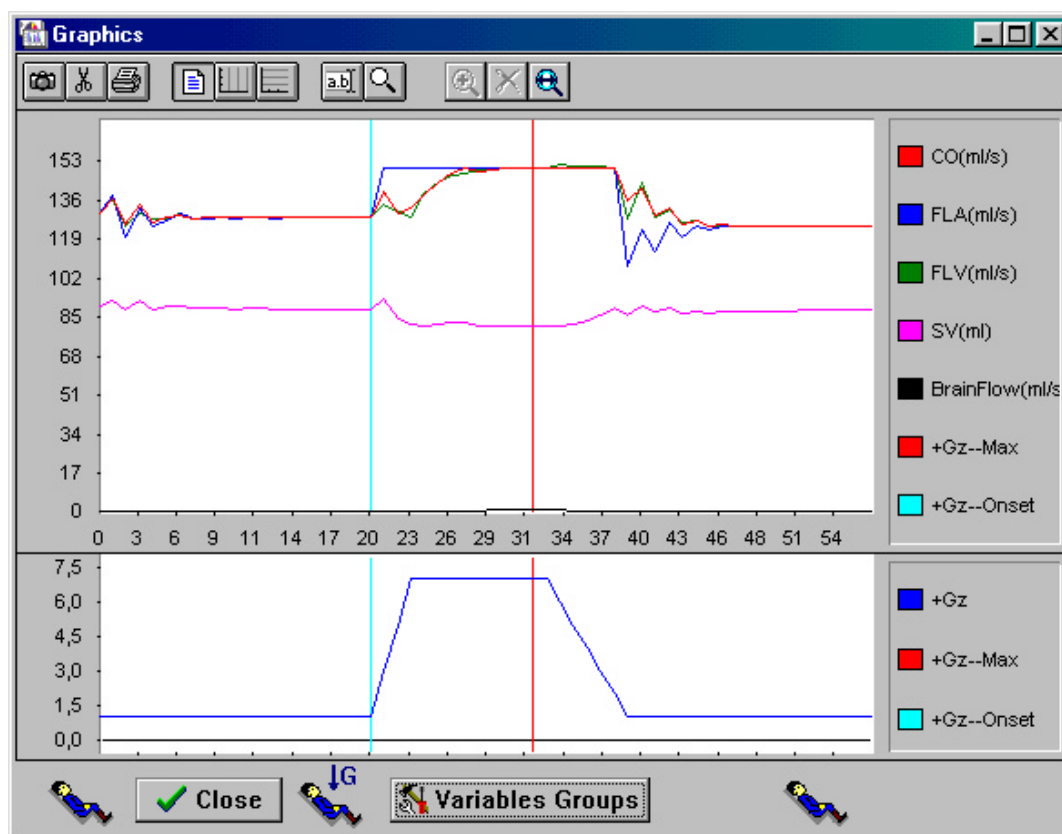


Fig.46B. The screen window illustrates an example of results' indication for the variables of Group4.

Save results of experiment in protocol

Experiment # 3 15.03.2002 14:15:29

Basic Model (regular flights)

Push-Pull effect
 Alpha = 1.00 Beta = 1.00 Gmax = 7.50 T = 10 T1 = 3
 Seatangles: A = 35° B = 55° C = -10°

Protections:
 Muscle Stress: Peripheral Straining (G-Threshold = 2.00)
 Anti-G Suit: Pneumatic
 G-Threshold = 2.00 Pressure Gradient = 1.51Psi (83 mm Hg)
 Abdomen covering = 20% Thigh covering = 55% Shank covering = 55%
 Breathing Pressure : Natural

Nervous Controls:
 Baroreflexes: Aortal ratio = 0.40 Carotid ratio= 0.60
 Vessels: NervContrRes Coef. = 1.00 NervContrD Coef. = 1.00
 NervContrU Coef. = 1.00
 Vessels Control Time Constant: 20.0
 Heart: NervContrHeartRate Coef. = 1.00 NervContrInotropism Coef. = 1.00
 Bainbridge's Reflex: Heart Rate Coef. = 1.00 Inotropism Coef. = 1.00
 Heart Control Time Constant: 4.0

Your comments
 You can write here your notes to experiment

OK Cancel Help

Fig.47. The screen form for saving the experiment results as a special Experiment protocol.

To get a hard copy of every experiment result as an experiment protocol (see Fig.47), one should first use the last right speed button. The “PILACCEL” gives the user an opportunity to write down their comments in this protocol (if this is deemed necessary) and save them together. After that one can print the protocol.

Experiment Parameters Setting

Models Acceleration Profiles Protections Interface options

Interface Options

Run-time warnings and confirmations

☒ Show steady-state regime warning

☒ "I can see again" message

☐ CLL automatic confirmation

☐ Ignore choice

☒ Cancel choice

☐ Abort choice

Time intervals for logging in sec 1.0

☐ Display process before steady-state

☐ Autosave protocol

Run-time controls

Time interval before loading 20

Time interval after loading 30

☐ Breath pulsation

OK Cancel

Fig.48. The screen form “Experiment Parameters Setting” in active state of “Interface Options”.

Fig.48 presents the screen form “Experiment Parameters Setting” in active state of “Interface Options”. This window serves settings of additional functions that may be individual preferable for different users. There are two blocks: 1. Interface Options; 2. Run-time controls. All functions that may be set using the appropriate positions are clearly enough described. Perhaps, two functions has to be additional described. The check-box “Breath pulsation” allows one to excecute a simulation experiment using variant of PM-model that reflects all pulsations of hemodynamic variables causing by sinusoidal breathing process. The check-box “Time intervals for logging in sec” allows the user to change the time density of graphics displaying.

4.3 How to use the incorporated Excel module.

As it was marked above, the “Pilaccel” software includes the Excel module that provides additional form presentation of results of simulation experiment. The graphic and table forms of results are available.

To use this presentation form, the user should push on appropriate speed-button or combination Ctrl+E on the keyboard. The following information window will appear.

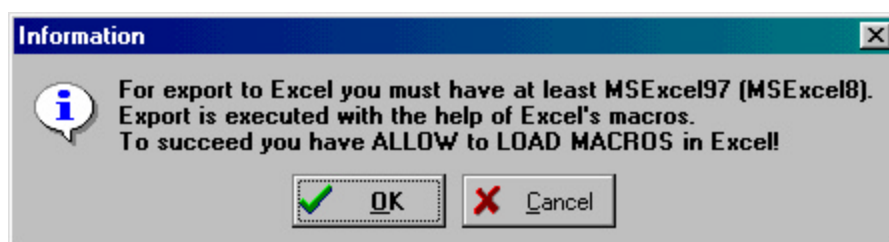


Fig.49. The information appearing every time the user is trying to export results of simulation experiment to incorporated Excel module and to use it.

If the appropriate software is set in computer, the user must follow the demands and push OK. The table and four groups of graphics will be available.